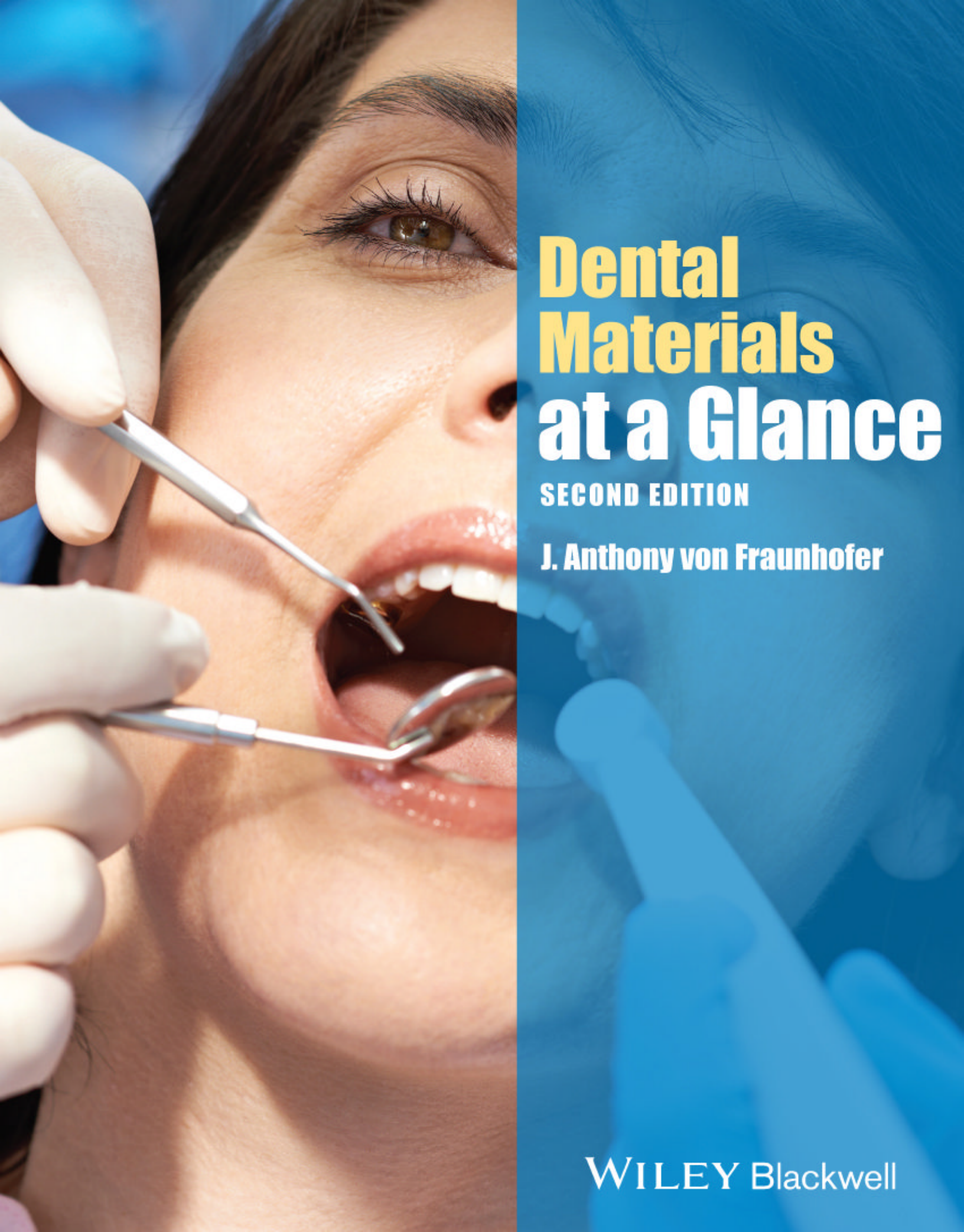




Dental Materials at a Glance

SECOND EDITION

J. Anthony von Fraunhofer

WILEY Blackwell

VISIT...

LANZAROTE
Caliente.COM

Dental Materials at a Glance

Second Edition

Dental Materials at a Glance

Second Edition

J. Anthony von Fraunhofer, BSc, MSc, PhD

Professor Emeritus

Baltimore College of Dental Surgery

University of Maryland

Maryland, USA

WILEY Blackwell

This edition first published 2013 © 2013 by John Wiley & Sons, Inc.
First Edition © 2010 J. Anthony von Fraunhofer

Editorial offices: 1606 Golden Aspen Drive, Suites 103 and 104, Ames, Iowa 50010, USA
The Atrium, Southern Gate, Chichester, West Sussex, PO19 8SQ, UK
9600 Garsington Road, Oxford, OX4 2DQ, UK

For details of our global editorial offices, for customer services and for information about how to apply for permission to reuse the copyright material in this book please see our website at www.wiley.com/wiley-blackwell.

Authorization to photocopy items for internal or personal use, or the internal or personal use of specific clients, is granted by Blackwell Publishing, provided that the base fee is paid directly to the Copyright Clearance Center, 222 Rosewood Drive, Danvers, MA 01923. For those organizations that have been granted a photocopy license by CCC, a separate system of payments has been arranged. The fee codes for users of the Transactional Reporting Service are ISBN-13: 978-1-1184-5996-6/2013.

Designations used by companies to distinguish their products are often claimed as trademarks. All brand names and product names used in this book are trade names, service marks, trademarks or registered trademarks of their respective owners. The publisher is not associated with any product or vendor mentioned in this book.

The contents of this work are intended to further general scientific research, understanding, and discussion only and are not intended and should not be relied upon as recommending or promoting a specific method, diagnosis, or treatment by health science practitioners for any particular patient. The publisher and the author make no representations or warranties with respect to the accuracy or completeness of the contents of this work and specifically disclaim all warranties, including without limitation any implied warranties of fitness for a particular purpose. In view of ongoing research, equipment modifications, changes in governmental regulations, and the constant flow of information relating to the use of medicines, equipment, and devices, the reader is urged to review and evaluate the information provided in the package insert or instructions for each medicine, equipment, or device for, among other things, any changes in the instructions or indication of usage and for added warnings and precautions. Readers should consult with a specialist where appropriate. The fact that an organization or Website is referred to in this work as a citation and/or a potential source of further information does not mean that the author or the publisher endorses the information the organization or Website may provide or recommendations it may make. Further, readers should be aware that Internet Websites listed in this work may have changed or disappeared between when this work was written and when it is read. No warranty may be created or extended by any promotional statements for this work. Neither the publisher nor the author shall be liable for any damages arising herefrom.

Library of congress cataloging-in-publication data

Von Fraunhofer, J. A. (Joseph Anthony), author.

Dental materials at a glance / J. Anthony von Fraunhofer. – Second edition.

p. ; cm. – (At a glance series)

Includes bibliographical references and index.

ISBN 978-1-118-45996-6 – ISBN 978-1-118-64648-9 (PDF) – ISBN 978-1-118-64664-9 (Pub) – ISBN 978-1-118-64666-3 (Mobi) – ISBN 978-1-118-68458-0 – ISBN 978-1-118-68461-0

I. Title. II. Series: At a glance series (Oxford, England)

[DNLN: 1. Dental Materials–Handbooks. WU 49]

RK652.5

617.6'95–dc23

2013007106

A catalogue record for this book is available from the British Library.

Wiley also publishes its books in a variety of electronic formats. Some content that appears in print may not be available in electronic books.

Cover design by Modern Alchemy LLC

Set in 9/11.5 pt Times by Toppan Best-set Premedia Limited

This book is dedicated to Dental Students, for they are the future of dentistry, and to the Faculty and Staff of Dental Schools because their expertise, dedication, and hard work make it all possible.

“Every tooth in a man’s head is more valuable than a diamond.”

Miguel de Cervantes, *Don Quixote* (1605)

Contents

Preface ix

Part I Fundamentals 1

- 1 Properties of materials—tensile properties 2
- 2 Toughness, elastic/plastic behavior, and hardness 4
- 3 Physical properties of materials 6
- 4 Adhesion and cohesion 8
- 5 Mechanical adhesion 10
- 6 Dental hard tissues 12
- 7 Bone 14

Part II Laboratory materials 17

- 8 Gypsum materials 18
- 9 Die materials 20
- 10 Dental waxes 22
- 11 Investments and casting 24

Part III Dental biomaterials 27

- 12 Inelastic impression materials 28
- 13 Elastic impression materials 30
- 14 Occlusal (bite) registration materials 32
- 15 Precious metal alloys 34
- 16 Base metal alloys 36
- 17 Porcelain bonding alloys 38
- 18 Implant metals 40
- 19 Partial denture base materials 42
- 20 Complete denture bases—acrylic resin 44

- 21 Modified acrylics and other denture base resins 46
- 22 Denture fracture and repair 48
- 23 Denture liners 50
- 24 Denture cleansing 52
- 25 Dental luting 54
- 26 Cavity varnishes, liners, and bases 56
- 27 Provisional (temporary) dental cements 58
- 28 Inorganic (acid–base reaction) cements 60
- 29 Resin-modified and resin cements 62
- 30 Denture adhesives 64
- 31 Dental amalgam 66
- 32 Adhesive dentistry 68
- 33 Bonding to dentin 70
- 34 Composite restorative resins 72
- 35 Endodontic filling materials 74
- 36 Provisional filling materials and restorations 76
- 37 Materials in periodontics 78
- 38 Dental porcelain 80
- 39 Manipulation and properties of porcelain 82
- 40 Strengthening porcelain 84
- 41 Advanced ceramic systems 86
- 42 CAD-CAM restorations 88
- 43 Orthodontic materials 90
- 44 Grinding, polishing, and finishing 92
- 45 Adverse effects of dental biomaterials 94
- 46 Dental erosion 96

Glossary 98

Index 101

Preface

Every effort was made to provide good coverage of each important and significant area of dental materials science in the First Edition. Since then, dental materials science has advanced and it became clear that a completely revised and greatly expanded second edition was necessary. Inevitably in this new edition, certain subjects have not been covered in depth and still others probably have not been discussed in the detail that many specialists might wish. Nevertheless, the essentials have been treated concisely and as completely as possible within a tight framework.

It must be stressed that the reader should understand that the coverage provided here cannot hope to rival that of the much larger and comprehensive standard texts in the field. Accordingly, the reader is encouraged to consult these texts, listed here, when there is a need for more detailed discussion and explanation.

Finally, this book is not intended to replace lectures and formal course work but rather to function as a concise guide and expanded revision notes to the large, complex, and continuously developing field of dental biomaterials science. Mention of standards and specifications has been made at various times and these are references to ADA/ANSI

and ISO specifications, which are readily available. Accordingly, specification details are not stated here.

On a personal note, I should like to express my appreciation of my wife Susan for her patience, support, and forbearance while I labored on this book. I must also express my appreciation of the advice, comments, and enthusiastic support of my many friends and colleagues. They are too many to mention here, but they know who they are and their input is greatly appreciated.

J. Anthony von Fraunhofer

Recommended standard texts

Applied Dental Materials, 9th edition. J.F. McCabe and A.W.G. Wells, Blackwell, Oxford, UK (2008).

Craig's Restorative Dental Materials, 13th edition, R.L. Sakaguchi and J.M. Powers (editors), Mosby-Elsevier, St. Louis, MO (2011).

Phillips' Science of Dental Materials, 12th edition, K.J. Anusavice (editor), Saunders-Elsevier Science, St. Louis, MO (2012).

Dental Materials at a Glance

Second Edition

Part I

Fundamentals

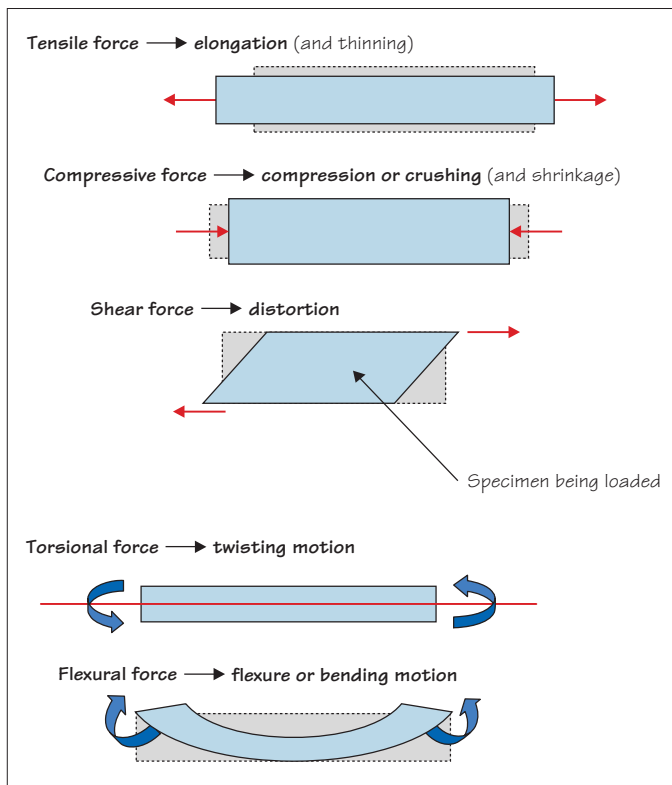


Figure 1.1 Applied forces and specimen deformations.

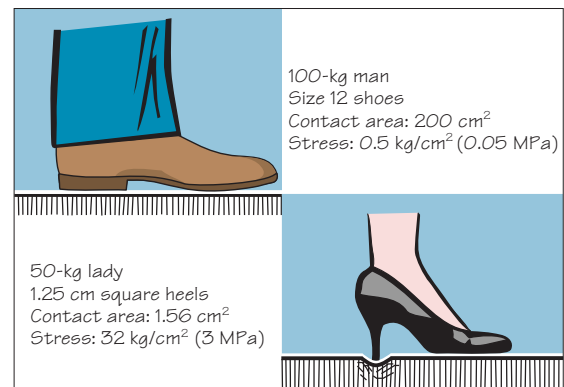


Figure 1.2 Load versus stress for feet.

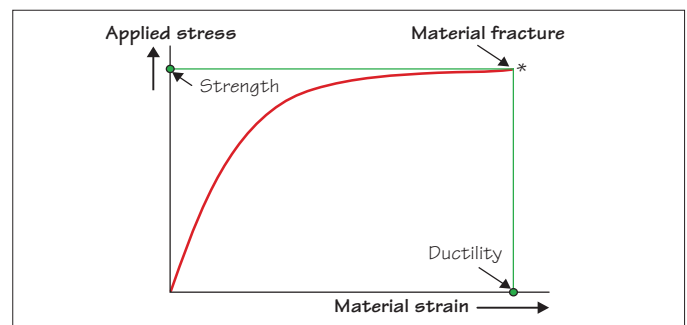


Figure 1.3 The stress-strain curve of a nonferrous metal.

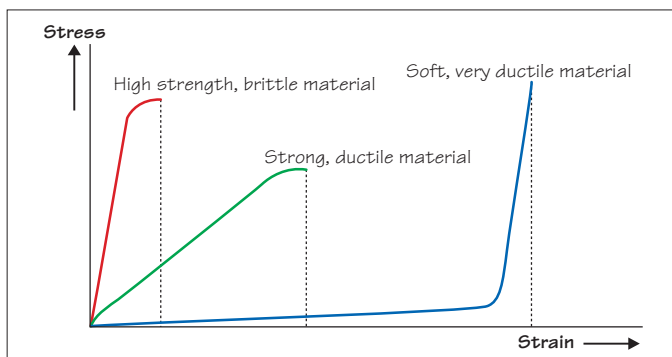


Figure 1.4 Stress-strain curves for brittle, elastic, and ductile materials.

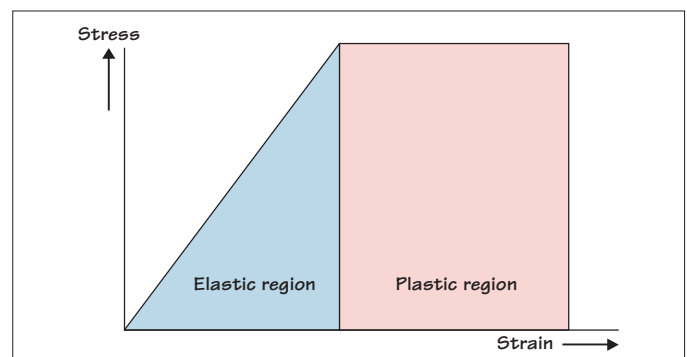


Figure 1.5 Elastic and plastic regions of a stress-strain curve.

Box 1.1 Desirable properties of dental materials

Biocompatibility
Absence of toxicity
Aesthetic appearance
Strength and durability
Low solubility
Ease of manipulation
Long shelf life
Simple laboratory processing
Long working time
Rapid/snap set

Table 1.1 Typical mechanical properties of dental biomaterials

Material	Tensile strength (MPa)	Compressive strength (MPa)	Shear strength (MPa)	Elastic modulus (GPa)	Hardness (KHN)
Gold alloy	448			77	22
Dental amalgam	54.7	318	188	34	110
Dentin	51.7	297	138	1.4	68
Enamel	10.3	384	90	4.6	343
Porcelain	24.8	149	111	140	460
Composite	45.5	237		14	
Zn phosphate cement	8.1	117	13	13.7	40
Die stone	7.7	48			
Ca(OH) ₂	1.0	10.3			
Glass ionomer	18	150		20	

Dental biomaterials are used in laboratory procedures and for the restoration and replacement of teeth and bone. Material selection must consider function, properties, and associated risks, and all dental biomaterials must satisfy certain criteria (Box 1.1).

Mechanical properties are important since teeth and restorations must resist biting and chewing (masticatory) forces. Typical material properties are given in Table 1.1.

Biting forces vary with patient age and dentition, decreasing for restored teeth and when a bridge, removable partial denture (RPD), or complete denture is present. Effects vary with the type of applied force and its magnitude. Types of applied force, and the resulting deformations, are shown in Figure 1.1.

1 Stress: σ , force per unit cross-sectional area

- Stress, the applied force and the area over which it operates, determines the effect of the applied load. For example, a chewing force of 72 kg (10 N) spread over a quadrant 4 cm² in area exerts a stress of 18 kg/cm² (1.76 MPa). However, the same force on a restoration high spot or a 1-mm² hard food fragment produces a stress of 7200 kg/cm² (706 MPa), a 400-fold increase in loading. This stress effect is one reason that occlusal balancing is essential in restorative dentistry. A more graphic example of the difference between applied force and stress is shown in Figure 1.2. This example also clearly indicates why it is more painful when a woman wearing high heels steps on you than when a man does!

2 Strength: The stress that causes failure

3 Ultimate strength: The maximum stress sustained before failure

4 Proportional limit: The maximum stress that the material can sustain without deviation from linear stress–strain proportionality

5 Elastic limit: Maximum stress that can be applied without permanent deformation

6 Yield strength: σ_y , stress at which there is a specified deviation from stress-to-strain proportionality, usually 0.1%, 0.2%, or 0.5% of the permanent strain

7 Strain: ϵ , ratio of deformation to original length, $\Delta L/L$; measures deformation at failure

8 Ductility: Percentage elongation, i.e. $\Delta L/L \times 100\%$

- **Ductile** materials exhibit greater percentage elongation than brittle materials and can withstand greater deformation before fracture.

9 Burnishing index: Ability of a material to be worked in the mouth or burnished, expressed as the ratio of % elongation to yield strength

10 Poisson's ratio: ν , ratio of lateral to axial strain under tensile loading; denotes reduction in cross-section during elongation

- **Brittle** materials have low ν values, i.e. little change in cross-section with elongation, whereas ductile materials show greater reduction in cross-section, known as specimen necking.

11 Elastic modulus: E , ratio of stress to strain, also known as **modulus of elasticity** or **Young's modulus**; denotes material stiffness and is determined as the slope of the elastic (linear) portion of the stress–strain curve

12 Stress–strain curves: Generated by applying a progressively increasing tensile force while measuring applied stress and material strain until fracture occurs

The shape of the stress–strain curve indicates the properties of the material (Figure 1.3 and Figure 1.4):

- Nonferrous metals (e.g., gold and copper) show a continuous curve to failure whereas ferrous materials exhibit a “kink” in the curve, known as the **yield point**.
- The intersection of a line parallel to the abscissa (strain) axis from the failure point to the ordinate (stress) axis is **specimen strength** whereas the vertical line from the failure point to the strain axis is the **ductility**.
- High-strength, brittle materials show steep stress–strain curves with little strain at failure, e.g. ceramics.
- Strong ductile materials, e.g. metals, show moderate slopes in the stress–strain curve but good extension until failure.
- Soft ductile materials, e.g. elastomers, show long, shallow linear stress–strain behavior followed by a sharp rise in the curve when, with increasing applied force, the elastomer no longer extends linearly (or elastically) and failure occurs.

13 Resilience: Resistance to permanent deformation (i.e., energy required for deformation to the proportional limit); given by the area under the elastic portion of the stress–strain curve (Figure 1.5)

14 Toughness: Resistance to fracture (i.e., energy required to cause fracture); given by the total area (i.e., both the elastic and plastic regions) under the stress–strain curve (Figure 1.5)

15 Hardness: Resistance to penetration; a measure of scratch resistance

- Hardness is measured by several techniques, including the **Barcol, Bierbaum, Brinell, Knoop, Rockwell, Shore, and Vickers tests**.

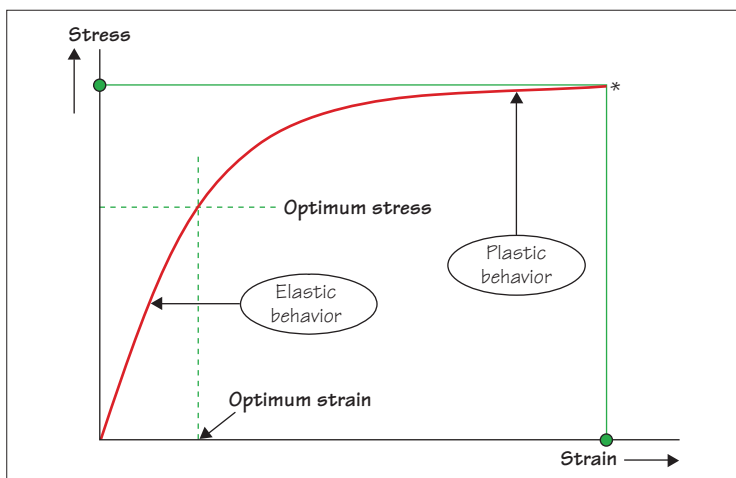


Figure 2.1 Optimal loading (stress and strain) region for a resilient material.

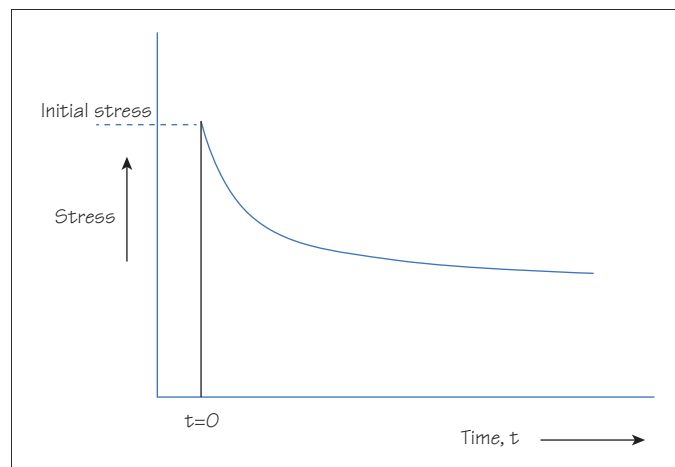


Figure 2.2 Stress relaxation: decrease in induced stress as a result of creep.

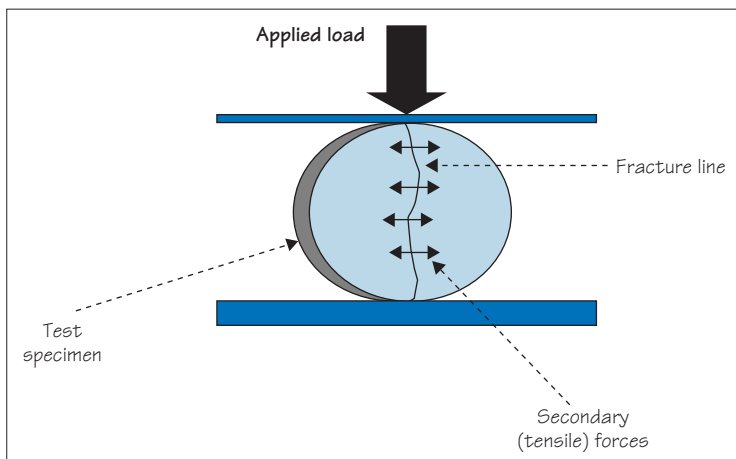


Figure 2.3 Diametral disc test for determining the tensile strength of brittle materials.

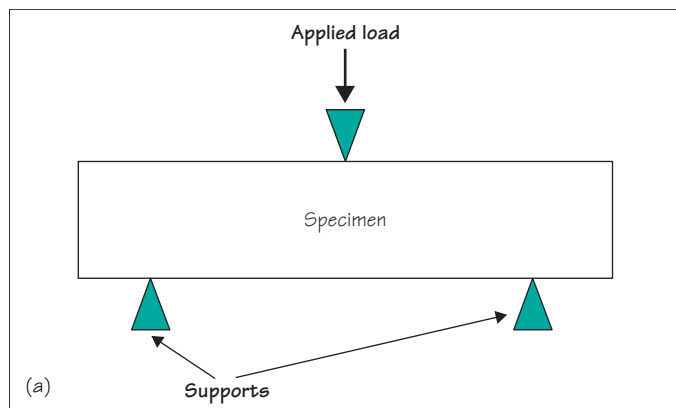


Figure 2.4a Transverse testing of a specimen.

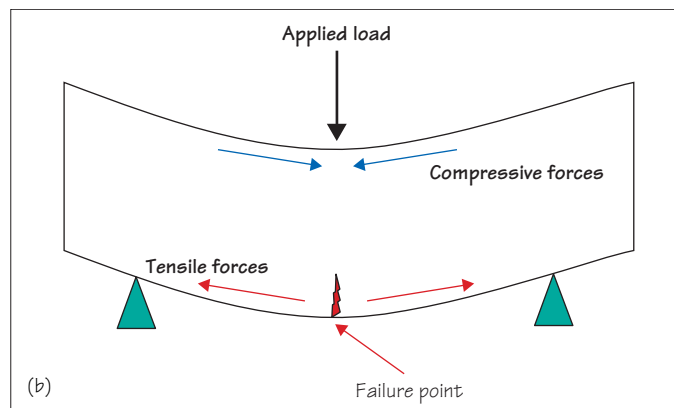


Figure 2.4b Loads and resultant stresses in a specimen under transverse testing.

2.1 Elastic and plastic behavior

Elastic materials deform (strain) instantaneously when loaded but, when the load is released, the specimen will resume its original dimensions although the recovery rate varies with the material. Deformation (strain) is directly proportional to the applied load (stress) in accordance with **Hooke's law** up to the proportional limit. Elasticity is usually the result of bond stretching along crystallographic planes in an ordered solid. Subjecting an elastic material to a load above its elastic limit will induce a degree of plastic (permanent) deformation. Ideally, applied loads should never exceed the elastic limit (Figure 2.1).

Plastic materials, typically **polymers** or **resins**, deform when loaded but the deformation is not proportional to the applied load—behavior known as nonlinear or non-Hookean deformation—due to their viscoelasticity. Upon release of the applied force, the specimen does not completely recover its original dimensions and is said to be **plastically deformed**.

2.2 Viscoelasticity

Viscous materials, e.g. honey, resist shear flow and show linear strain over time under an applied stress, i.e. time-dependent strain due to diffusion of atoms or molecules inside an amorphous material. In contrast, elastic materials deform instantaneously when loaded. Materials that exhibit both viscous and elastic characteristics when deforming are described as **viscoelastic**.

2.3 Stress relaxation

Polymers are viscoelastic, exhibiting both elastic and plastic behavior, as well as time-dependent strain. When polymers are subjected to constant load, they undergo continuing strain over time, known as **creep**, and the stress experienced by the polymer decreases, an effect known as **stress relaxation**. In other words, the stress induced in the specimen decreases over time (Figure 2.2).

2.4 Fracture toughness

Fracture toughness is the ability to deform plastically without fracture and is proportional to energy consumed in plastic deformation. Cracks or flaws, arising naturally or developing over time, cause weakening such that fracture may occur at stresses below the yield stress, the flaw acting as a **stress riser**.

Flaws cause problems because brittle materials under loading cannot deform plastically and redistribute stresses. As the flaw or crack size increases, the stress for specimen failure decreases. This behavior is expressed by the stress intensity factor, K , which is determined by the stress and the crack length. Fracture occurs when the stress intensity reaches a critical value, K_c , given by $Y \cdot \sigma \cdot \sqrt{\pi a}$, where Y is a function of crack size and geometry, and a is the crack length.

This critical value is known as the **fracture toughness** of the material.

2.5 Determining mechanical properties

1 Tensile properties: Discussed in Chapter 1; measured on flat specimens with a “necked” region or on dumbbell-shaped specimens

Brittle materials (e.g., amalgam and ceramics) cannot be tested in tension and their tensile properties are determined by the **diametral tensile test**. In testing, a compressive load (P) is applied to a vertical disc of material and induces a tensile force along the specimen diameter (Figure 2.3). The diametral tensile strength (DTS) is given by

$$\text{DTS} = \frac{2P}{\pi \cdot (\text{diameter}) \cdot (\text{thickness})}$$

2 Compressive strength: Determined by applying a compressive load to a cylindrical or square cross-section specimen; expressed as the load to failure divided by cross-sectional area

3 Shear strength: Determined by applying a tensile stress to a lapped specimen, by a modified cantilever test or a pin-disc system; important when shear loading occurs, e.g., with veneers

4 Transverse strength: Measured in a specimen of length L supported at the ends with a load (P) applied in the middle (Figure 2.4a, Figure 2.4b)

Transverse failure initiates at the lower edge where the applied force induces tensile stresses while compressive forces occur in the upper region. Strength is given by stress at failure:

$$\text{Stress } \sigma = \frac{3PL}{2 \cdot (\text{width}) \cdot (\text{thickness})^2}$$

$$\text{Deformation } \delta = \frac{P \cdot L^3}{4 \cdot E \cdot (\text{width}) \cdot (\text{thickness})^3}$$

where E is the modulus. Transverse strength is important for denture bases.

5 Indentation hardness: Resistance to penetration, determined by measuring the indentation produced in the specimen by an indenter under load

The most important hardness tests in dentistry are the Knoop and Shore tests:

Knoop hardness test: The test uses a nonsymmetrical diamond point (7:1 ratio of length to width) and the Knoop hardness number $\text{KHN} = L/l^2 \cdot C_p$, where L is the applied load, l is the length of the long diagonal, and C_p is a constant that relates l to the indentation area; the test requires a flat, highly polished specimen but no load is specified so it can be used on a microscopic scale for both ductile and brittle materials.

Shore hardness test: This test measures penetration of a blunt indenter into a soft or elastic material and is useful for soft materials, e.g. elastomeric materials.

Hardness values can provide an indication of the resistance of materials to scratching, wear, and abrasion.

2.6 Abrasion and wear resistance

Abrasion and wear are important for polymeric restorations, for ceramic restorations opposing natural teeth, and for dentifrices. Surface hardness is not always a reliable guide to wear resistance, particularly for hard, brittle materials or for elastomers. Various abrasion/wear test systems are used, the simplest being reciprocating arm abraders with nylon brushes or rubber cups mounted on counterbalanced arms driven over the test piece. Weights placed on the arm vary the applied load while water, artificial saliva, or dentifrice slurries can be applied to the test piece surface. More complex test arrangements have specimens mounted on or subjected to rotating or oscillating heads, again with abrasives applied to the test specimen surface. Wear/abrasion damage is assessed by profilometry (change in the surface profile), weight loss, or both. No abrasion system completely mimics behavior in the oral cavity and both data quantification and reproducibility can present problems. Nevertheless, abrasion/wear testing can provide useful predictive data with regard to material performance.

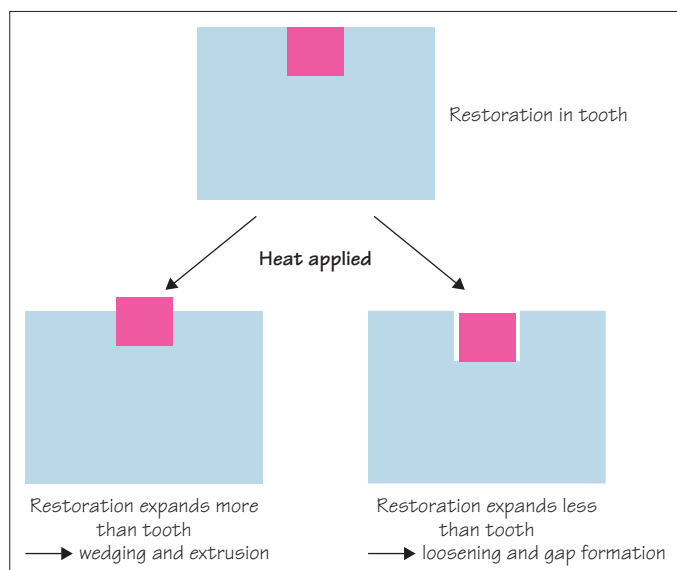


Figure 3.1 Effect of temperature rise on a restoration and tooth with different coefficients of thermal expansion.

Table 3.1 Thermal properties of various dental materials

Material	Thermal conductivity (J/s/cm ² /°C/cm)	Specific heat (J/g/°C)	Thermal diffusivity (mm ² /s)
Copper	3.84	0.38	
Gold	2.97	0.13	119.0
Mercury	0.084	0.14	
Platinum	0.698	0.13	
Silver	4.21	0.23	
Dental amalgam	0.23		9.6
Zinc phosphate cement	0.012		0.290
Zinc oxide–eugenol cement	0.005		0.389
Acrylic resin	0.002	1.46	0.123
Composite resin	0.011		0.675
Porcelain	0.010	1.09	0.64
Enamel	0.0092	0.75	0.469
Dentin	0.0063	1.17	0.18–0.26
Water	0.0044	1.00	0.14

Table 3.2 Coefficients of thermal expansion

Material	Coefficient of thermal expansion (×10 ⁻⁶ /°C)
Tooth (crown portion)	11.4
Amalgam	22.1–28.0
Gold	14.4
Composite resin	17–50
Acrylic resin	76.0
Porcelain	12.0
Glass ionomer	10.2–11.4
Inlay wax	350–450
Silicone impression material	210
Polysulfide impression material	140

Table 3.3 Electrical constants for dental materials and teeth

Material	Resistivity (Ω·cm)	Dielectric constant
Tooth enamel	2.6–6.9 × 10 ⁶	
Dentin	1.1–5.2 × 10 ⁴	8.6
Glass ionomer	0.8–2.5 × 10 ⁴	2–7 × 10 ⁵
Zinc oxide–eugenol	10 ⁹ –10 ¹⁰	10
Zinc polyacrylate	0.4–4 × 10 ⁵	4 × 10 ³ –2 × 10 ⁵
Zinc phosphate	2 × 10 ⁵	

Table 3.4 Wavelengths of visible light

Color	Approximate wavelength interval (nm)
Red	630–700
Orange	590–630
Yellow	560–590
Green	490–560
Blue	450–490
Indigo	420–440
Violet	400–450

Physical properties relevant to dental biomaterials include thermal, electrical, and optical properties.

3.1 Thermal and electrical properties

Typical thermal parameters are given in Table 3.1.

1 Thermal conductivity: K , the rate of heat conduction through a unit cube of material for a temperature difference of 1°C across the cube, expressed in $\text{J/s}\cdot\text{cm}^2/^\circ\text{C}/\text{cm}$ ($\text{J}\cdot\text{s}^{-1}\cdot\text{cm}^{-2}\cdot^\circ\text{C}^{-1}\cdot\text{cm}^{-1}$)

- Metal restorations have higher K values than teeth and cause greater pulp temperature changes than hard tissue during exposure to hot or cold liquids.

2 Specific heat: C_p , the quantity of heat that raises the temperature of 1 g of substance by 1°C , expressed in $\text{J/g}/^\circ\text{C}$ ($\text{J}\cdot\text{g}^{-1}\cdot^\circ\text{C}^{-1}$)

- Specific heat determines the heat input required to reach the metal's melting point during casting. C_p is lower for gold than for nonprecious and base metal alloys, and the latter require greater heat input to melt than gold.

3 Thermal diffusivity: Δ , defined as $K/C_p \times \rho$ (i.e., thermal conductivity divided by specific heat multiplied by the density), expressed in mm^2/s ($\text{mm}^2\cdot\text{s}^{-1}$)

- Diffusivity characterizes transient heat flow, determining the rate at which a material approaches thermal equilibrium; it accounts for the thermal shock to the pulp found with metallic restorations.

4 Lining efficiency: Z , the thermal protection by liners; determined by $Z = T/\sqrt{\Delta}$, where T is the liner thickness

5 Linear coefficient of thermal expansion: α , change in length per unit length of material for 1°C change in temperature, expressed as “ $^\circ\text{C}$ ” ($^\circ\text{C}^{-1}$) or sometimes as parts per million (ppm)

- Typical values are given in Table 3.2; α is temperature- and state-dependant, changing at the glass transition temperature (T_g) for polymers (see later).
- If expansion coefficients of restorations and tooth differ markedly, the relative expansions and contractions may result in gap formation and leakage (Figure 3.1). The high α value of waxes compensates for the shrinkage of dental alloys when casting restorations.

6 Electrical conductivity (κ , $\text{ohm}^{-1}\cdot\text{cm}$) and resistivity (ρ , $\text{ohm}\cdot\text{cm}$): Conductance $L = \kappa(A/l)$ whereas resistance $= \rho/(lA)$, where A is cross-sectional area and l is the length; conductance is the inverse of resistance. Resistivity values are given in Table 3.3.

- Dentin has a lower resistivity than enamel whereas sound enamel and carious enamel differ in resistivity. The conductivity of restorative materials may affect insulation by bases beneath metallic restorations.

7 Dielectric constant: ϵ , a measure of electrical insulation

- The high ϵ values for glass ionomer and polyacrylate cements indicate their ionic content, and the value of ϵ decreases as wet dental cement dries.

3.2 Optical properties (color and appearance)

Ideally, a restoration will match the natural hard and soft tissues but color is only partially inherent to a material because it is produced in

various ways, including selective reflection and absorption, scattering, diffraction, and interference. Thus a specimen's color is determined by composition, thickness, surface roughness, and the incident light. Further, the apparent color and light reflectance will vary with the background upon which the material is viewed.

Visible light perceived by humans has wavelengths in the range of 400–700 nm (Table 3.4). In fact, humans are **trichromatic**, with three types of color receptors: short-wavelength (**S cones**), most sensitive to violet (420 nm) light; middle-wavelength (**M cones**), most sensitive to green (534 nm) light; and long-wavelength (**L cones**), most sensitive to yellow-green (564 nm) light. Although humans can distinguish up to 10^7 different colors, the eye's receptor cones reduce the wavelengths of light to three color components known as **tristimulus values**. Further, because of this human trichromaticity, the perception of a spectral color may alter with its intensity.

A material's color is frequently measured by the **CIE** (Commission International de l'Eclairage) system, which defines color by three parameters: L^* , a^* , and b^* . The brightness or **value** (L^*) denotes the lightness or darkness of a color whereas the dominant wavelength (**hue**) is its direction from white in a color wheel or chromaticity diagram. The CIE system represents this by the relative values of a^* and b^* and their signs: $-a^*$ denotes increasing greenness whereas $+a^*$ denotes increasing redness; $-b^*$ denotes an increase in blue-violet and $+b^*$ denotes an increase in yellow-green. The intensity (**chroma**) of a color and its purity is represented by the distance from the center of the chromaticity diagram, i.e. by the magnitudes of the values of a^* or b^* .

Color may also be determined by the **Munsell color system**, in which it is compared with a large number of color tabs. **Value** (lightness) is determined first over a range of 10 for white to 0 for black followed by determination of **chroma**, ranging from 0 for gray to 18 for highly saturated color. Finally, **hue** is determined by matching with color tabs of the determined value and chroma. For this, hue is measured on a scale of 2.5 to 10 for each color family, namely red, yellow red, yellow, etc. Thus a color may be specified as 4R 7/3, indicating a hue of 4R, a value of 7, and a chroma of 3. Colors specified using the Munsell system can be compared using a color difference calculation that quantifies differences detected by trained observers.

Metamerism is a phenomenon that can cause problems in color matching; metameric colors have the same tristimulus values under one light source but differ in their spectral energy distributions so that they may match in one light but not under others. Since the dominant light wavelengths of artificial and sunlight differ, color matches between restorations and teeth vary with the incident light, complicating satisfactory matching of teeth and restorations. Ideally, shade selection/matching is performed under conditions that reproduce use.

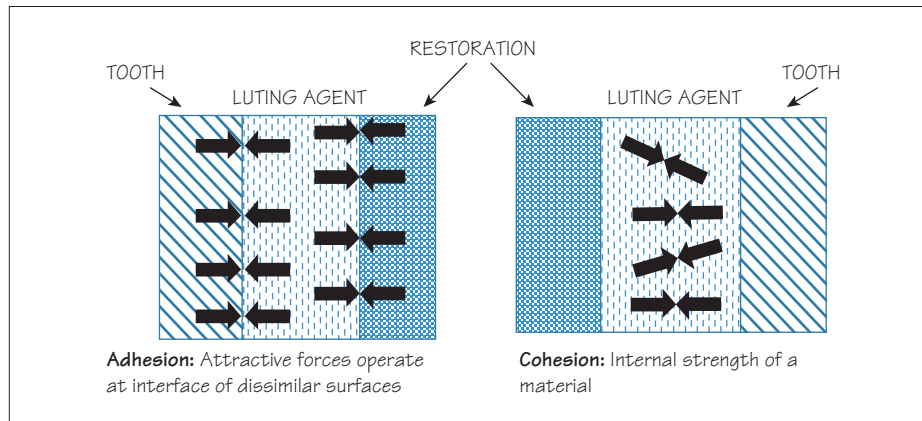


Figure 4.1 Adhesion and cohesion.

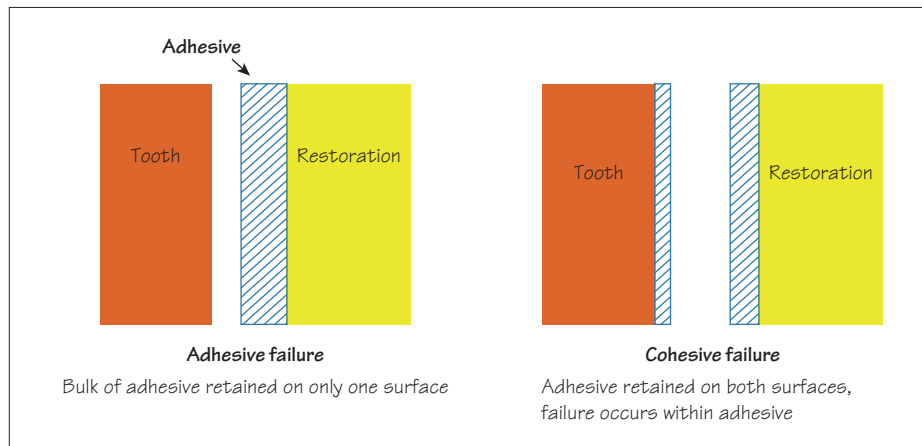


Figure 4.2 Adhesive and cohesive failures of cemented restorations.

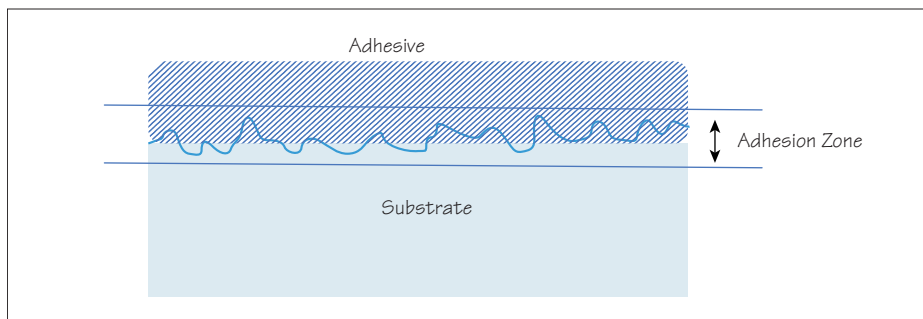


Figure 4.3 Adhesion zone between adhesive and substrate (schematic).

Box 4.1 Molecular forces determining cohesive strength of an adhesive

- 1 The chemical bonds within the adhesive material
- 2 Chemical bonds due to cross-linking of the polymer(s) within a resin-based material
- 3 Intermolecular interactions between the adhesive molecules
- 4 Mechanical bonds and interactions between the molecules in the adhesive

Table 4.1 Basic types of adhesion

Type	Characteristics
Specific	Molecular attraction between contacting surfaces
Mechanical	Adhesion through mechanical interlocking between adhesive and substrate surfaces
Effective	Bonding between adhesive and substrate due to a combination of specific and mechanical adhesion

Adhesion is the molecular attraction between the contacting surfaces of dissimilar molecules whereas **cohesion** is the molecular attraction uniting similar molecules throughout a material. Adhesion binds an adhesive to the substrate whereas cohesion binds the individual components of the adhesive, i.e. the internal strength of an adhesive (Figure 4.1). Both adhesion and cohesion determine overall bonding effectiveness; a bonded restoration will fail if the luting agent separates from the substrate (adhesive failure) or there is internal breakdown of the adhesive (cohesive failure), shown in Figure 4.2.

4.1 Forces in cohesion

A number of molecular forces determine the cohesive strength of an adhesive (Box 4.1) and affect the properties of the adhesive, notably consistency, flow, and viscosity. During setting, solidification occurs through intermolecular bonds within the adhesive, by formation of new bonds and by strengthening of existing bonds, typically cross-linking of short-chain molecules to form longer chains and/or three-dimensional networks of molecular chains. Thus, an adhesive's cohesive strength is affected by the curing conditions and when cured under suboptimal conditions, the adhesive lacks cohesive strength.

4.2 Forces in adhesion

Adhesion can be divided into three basic types (Table 4.1). Specific adhesion, due to molecular interactions between adhesive and substrate, can be divided into three different types: chemical, dispersive, and diffusive adhesion. Micromechanical effects also can be involved in the overall adhesion phenomenon through the adhesive attaching to a roughened substrate and augmenting adhesion; see Chapter 5.

4.3 Mechanisms of adhesion

4.3.1 Chemical adhesion

If the adhesive and substrate can form a compound at their interface, the developing ionic or covalent bonds result in a strong bond (chemical adhesion) between the two but adhesion is weaker when there is only hydrogen bonding. The lower chemical adhesion with hydrogen bonds is because despite having comparable lengths to covalent and ionic bonds, they are an order of magnitude weaker. The strengths of chemical bonds can be high (Table 4.2), their lengths are short, and, for good adhesion, the surfaces must remain in proximity for a stable bond.

Chemical adhesion is uncommon in dentistry confined to reactions between carboxylate-based luting agents and calcium in hard tissues. When present, it can account for $\leq 50\%$ of all interactions although long-term stability depends on resistance to moisture.

4.3.2 Dispersive adhesion

In dispersive adhesion (physisorption), the material surfaces are held together by **van der Waals forces**, attractive forces between two molecules, each of which has region(s) of small positive and negative

Table 4.2 Bond energies and bond lengths in adhesive forces

	Average bond energy (kJ/mol)	Average bond length (nm)
Ionic bond	600	0.25
Covalent bond	550	0.15
Metallic bond	250	0.40
Hydrogen bond	25	0.20
Van der Waals forces	8.5	0.45

charge such that the molecules are polar with respect to the molecule's average charge density. If these positive and negative poles are inherent to the molecule, they are known as **Keesom** forces, whereas if the polarity is a transient effect due to random electron motions, the forces are known as **London** forces. London dispersion forces are useful in adhesion because they arise without the need for permanent polarity in either adhesive or substrate.

Although van der Waals bond lengths are relatively long (Table 4.2), the forces only act over very small distances. About 99% of the work required to break van der Waals bonds is performed once the joined surfaces are separated by more than a nanometer; i.e. the effectiveness of adhesion due to chemical or dispersive bonding is limited. Once a crack is initiated, it propagates easily along the interface because of the brittleness of the interfacial bonds and, consequently, greater contact surface areas often have little effect on adhesion.

4.3.3 Diffusive adhesion

Some materials may merge (intermingle) at the bonding interface by diffusion, typically when their molecules are mobile and/or soluble in each other. This form of interaction or **interdigitation** occurs when a resilient denture liner is processed onto an acrylic resin denture base. In the former, bonding arises from the mutual solubility and interactions between monomer in the liner material and the denture surface of the acrylic base, with diffusive adhesion arising from interdigitation of polymer chains. However, mobility of the polymer molecules influences their interdigitation and diffusive bonding. Thus the restricted mobility of cross-linked polymers limits diffusion and interdigitation compared with more mobile and better interdigitating non-cross-linked polymers.

Diffusive adhesion is also involved in sintering, e.g. firing porcelain to a metal surface during fabrication of a PFM crown. Since diffusive adhesion requires interaction of atomic species between two surfaces, the longer the interaction between the two surfaces, the more diffusion occurs and, accordingly, the stronger the adhesion.

4.4 The adhesion zone

The adhesive bonded to a substrate has a modified molecular structure at the bonding interface. This interfacial region or **adhesion zone** (Figure 4.3), is characterized by the changes in the adhesive (and sometimes in the substrate) arising from bonding interactions.

The adhesive's chemical, mechanical, and optical properties differ from the bulk material in the adhesion zone; the latter varies in thickness, from nanometers up to a few millimeters, depending on the substrate surface, the composition and characteristics of the adhesive, and the curing conditions. Bond strength, for example, may be impaired because of inadequate cohesion within the adhesive. Such considerations affect the selection of the optimum luting agent for restorations.

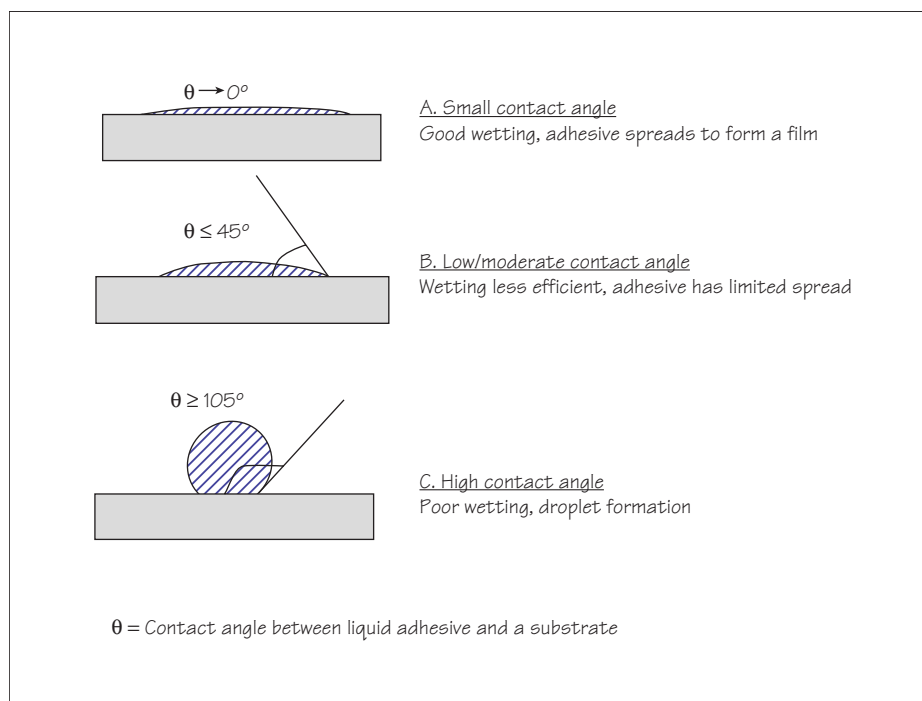


Figure 5.1 Adhesive-substrate contact angles for clean, slightly contaminated, and contaminated surfaces.

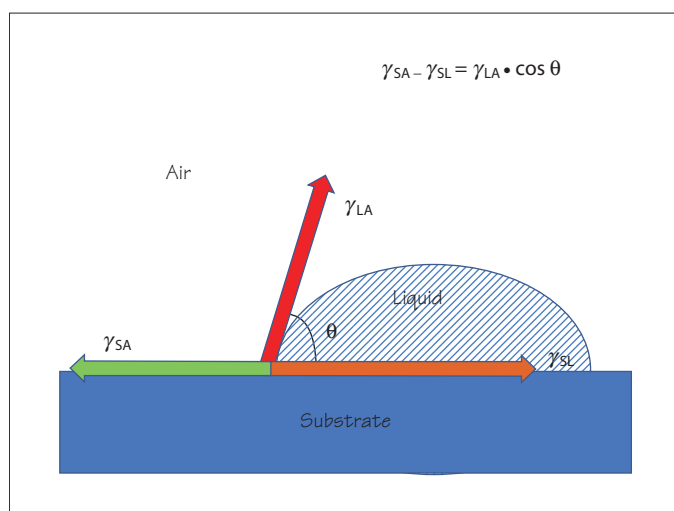


Figure 5.2 Interfacial tensions for a drop of liquid on a surface: θ , contact angle between liquid and substrate; γ_{SA} , solid-air interfacial tension (surface energy of substrate); γ_{LA} , liquid-air interfacial tension (surface tension of liquid); γ_{SL} , solid-liquid interfacial tension (adhesion between liquid and solid).

Table 5.1 Relationships among surface wetting, substrate surface energy, contact angles, and adhesive/cohesive forces

Surface	Substrate	Contact angle	Forces
Wetted	High energy	$\theta < 90^\circ$	Adhesion > cohesion
Poor/nonwetting	Low energy or contaminated	$\theta > 90^\circ$	Cohesion > adhesion

Mechanical (actually micromechanical) effects can significantly impact the bond between an adhesive and a substrate, particularly when contributions from chemical, dispersive, and diffusive adhesion are limited or absent. However, for mechanical adhesion to operate, the adhesive must wet the substrate and this is affected by the surface tension of the unset adhesive and the contact angle between it and the substrate.

5.1 Contact angle and surface tension

Adhesives that wet the substrate have larger contact areas than non-wetting materials, with wetting depending on the relative surface energies of adhesive and substrate. Low surface energy materials such as poly(tetrafluoroethylene), or PTFE, and silicone materials do not wet and resist adhesive bonding without special surface preparation.

The ability of a liquid to form an interface with a solid surface, i.e. the degree of wetting, is evaluated as the **contact angle** θ between the liquid and the substrate surface, θ being determined by both the liquid's surface tension and the nature and condition of the substrate surface. The degree of wetting increases with smaller contact angles and lower surface tensions (Figure 5.1). Good wetting occurs with clean surfaces, i.e. θ is close to 0° (Figure 5.1A), but the contact angle is greater, i.e. $0 < \theta < 90^\circ$, with a slightly contaminated surface (Figure 5.1B). With contaminated or low surface energy substrates, $\theta > 90^\circ$ (Figure 5.1C), a condition sometimes termed dewetting, and the liquid forms droplets on the substrate surface.

The contact angle θ is a function of both the adhesion between adhesive and substrate and the cohesion within the liquid adhesive. If there is strong adhesion to the substrate surface and weak cohesion within the liquid, there is a high degree of wetting (termed **lyophilic** conditions). Conversely, weak adhesion and strong cohesion, i.e. **lyophobic** conditions, results in high contact angles and poor wetting of the substrate surface. A small contact angle indicates a greater overall substrate surface energy and a high interactive force between the liquid and the substrate, resulting in greater adhesion due to a larger contact area between adhesive and substrate. For a low-energy or contaminated surface, $\theta > 90^\circ$ and cohesion within the adhesive can exceed the adhesion between liquid and substrate, with poor wetting (Table 5.1).

Surface scientists refer to **interfacial effects**, using the terms liquid–air interfacial tension γ_{LA} (i.e., the liquid surface tension), solid–liquid interfacial tension γ_{SL} (i.e., the surface tension between substrate and adhesive, which approximates the surface adhesion between liquid and solid), and the solid–air interfacial tension γ_{SA} (i.e., the surface tension between the solid and air, which approximates the surface energy of the solid), illustrated in Figure 5.2.

For a contact angle of θ° , these entities are related by **Young's equation**:

$$\gamma_{SA} - \gamma_{SL} = \gamma_{LA} \cdot \cos \theta$$

With complete wetting of the substrate, i.e. when $\theta = 0$ and $\cos \theta = 1$, Young's equation indicates that $\gamma_{LA} = \gamma_{SA} - \gamma_{SL}$ or $\gamma_{LA} \leq \gamma_{SA}$. In other words, if the adhesive surface tension (γ_{LA}) is less than the substrate surface energy (γ_{SA}), the adhesive will spread over the substrate.

The value of γ_{SA} when $\cos \theta = 1$ is the **critical surface energy** (CSE) and equals the value of γ_{SL} when the liquid just spreads over the

surface. The CSE of PTFE is 19.4 mJ/m^2 and the contact angle for water is 109.2° whereas these two parameters are 37.5 mJ/m^2 and 70.9° for acrylic resin, clearly indicating their differences in wetting (and bonding) behavior.

Wetting occurs when the adhesive surface tension (γ_{SL}) is less than the critical surface energy. This is often expressed as the **adhesion quotient** which requires the substrate surface energy (γ_{SA}) to exceed the surface tension of the adhesive liquid (γ_{SL}) by 10 dyne/cm . If the reverse is true, i.e. $\gamma_{SL} \geq \gamma_{SA}$, surface wetting is poor, adhesion is reduced because the adhesive tends to pull away from the surface during the curing process.

5.2 Micromechanical adhesion

When uncured, adhesives are fluid and can flow over the substrate, filling the voids, rugosity, and pores on the surface, attaching to that surface by mechanical interlocking. This is often referred to as micromechanical adhesion; see Figure 4.4.

Luting of restorations to teeth with dental cements primarily involves **micromechanical adhesion**, which probably also contributes significantly to bonding with resin-based adhesives as, for example, with fissure sealants and restorative resins. The effectiveness of micromechanical adhesion is largely determined by the luting agent wetting of the substrate in that poor wetting inhibits good apposition of cement and substrate. Further, the luting agent must be able to flow into the surface voids etc.; for this to occur, the adhesive must have a low viscosity. Water, for example, has a viscosity of $1 \text{ centipoise (cP)}$ and that of alcohol is 1.2 cP . Many other fluids have much higher viscosities, e.g. 9.22 cP for eugenol (oil of cloves), 1490 cP for glycerin, and $\sim 10^4 \text{ cP}$ for honey; the large viscosity difference between honey and water explains why the latter flows far more readily. It should be noted that the SI units for viscosity are pascal seconds (Pa-s) but are numerically equivalent in magnitude to cP values.

Inevitably, micromechanical adhesion of an adhesive to a surface is not simply a matter of wetting (i.e., contact angles) and the **rheological** (flow) properties of the adhesive. Other factors such as electrostatic forces (both attractive and repulsive) that may be operating between the adhesive, the substrate microtopography as well as a property of the applied fluid known as **thixotropy** affect micromechanical adhesion. A thixotropic fluid is one that under the action of mechanical forces such as stirring, vibration, or shear will temporarily transform to a state that has a lower viscosity with better flow than when it is in its static state. Thixotropic behavior is an important characteristic for endodontic (root canal) sealants, which are required to flow into a root canal, often under vibration. Further, thixotropy is often incorporated into paints by additives such as silicic acid and is probably present in various dental adhesive and cement formulations.

Thixotropy in an adhesive provides certain advantages to the overall adhesion system, particularly when a thixotropic adhesive is applied to a substrate because it will remain in place, even on vertical surfaces. Further, because adhesive flow is determined in part by the mechanical forces imposed during placement, there can be greater control of the adhesive film thickness combined with improved flow into the microtopography of the substrate surface.

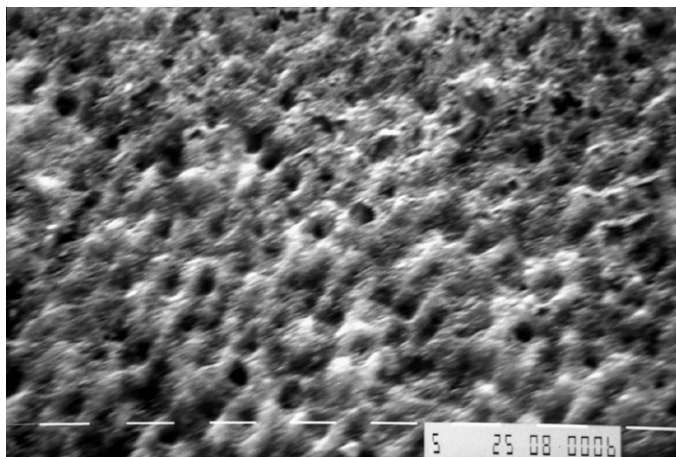


Figure 6.1 Scanning electron micrograph (x1000) of etched enamel.

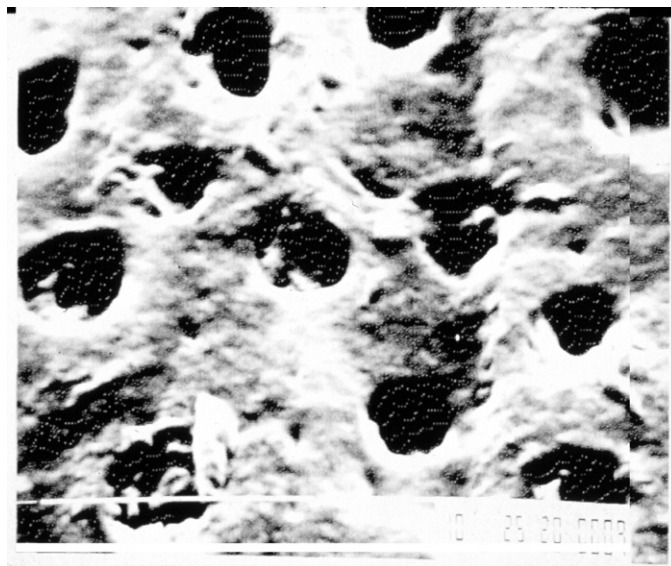


Figure 6.2 Scanning electron micrograph (x1000) of dentin tubules with smear layer present.

Table 6.1 Characteristics of dental hard tissues

	Enamel	Dentin	Cementum	Bone
Embryology	Ectoderm	Neural crest	Neural crest	Mesoderm but neural crest in the head
Formative cells	Ameloblasts	Odontoblasts	Cementoblasts	Osteoblasts
Degradative cells	Odontoclasts	Odontoclasts	Odontoclasts	Osteoclasts
Apatite content (%)	95	70	50	67
Ca/P ratio	1.66/0.03	1.97/0.06	Ca. 1.97	1.97/0.02
Organic matrix	Enamel proteins	Type I collagen and ground substance	Type I collagen and ground substance	Type I collagen and ground substance
Viability	No repair or remodeling	No remodeling; repair through 2° and 3° dentin	No remodeling; repair by deposition of new cementum	High remodeling rate; high potential for repair

Table 6.2 Average inorganic constituents content (wt.%) of mineralized tissue

Constituent	Enamel	Dentin	Bone
Ca	36	37	35
P	18	18	15
CO ₃	4	6	5
Na	0.6	0.4	0.5
Mg	0.2	1.2	0.5
F	0.01		
Mean Ca/P ratio	2.0	2.05	2.3

Table 6.4 Characteristics of dentinal tubules

	Near DEJ	Near pulp
Density (tubules/mm ²)	20,000	50,000
Diameter (μm)	0.5	2.5

Table 6.3 Inorganic and organic content (wt.%) of enamel and dentin

	Enamel	Dentin
Mineral (hydroxyapatite) content	96	70
Organic matter	1	20
Water	3	10

Box 6.1 Changes in dental enamel with patient age

Permeability decreases.
 Water content decreases.
 Surface composition changes through ion exchange with oral environment (e.g., fluoridation of enamel surface).
 Color darkens, in part through addition of organic matter to the enamel and sclerosis and staining of underlying dentin.
 Wear facets occur in areas of heavy function.

Box 6.2 Proteinaceous components of dentin

Collagenous proteins

(mainly Type I collagen with smaller amounts of Type V and Type I trimer collagens)

Noncollagenous dentin-specific proteins

(phosphophoryns, dentin sialoprotein, and dentin matrix protein-1)

Nonspecific proteins associated with mineralized tissues

(e.g., osteocalcin and osteopontin)

Dental hard tissue characteristics are summarized in Table 6.1, and Table 6.2 indicates mineral compositions.

6.1 Dental enamel

Enamel, covering the tooth crown, is highly mineralized (Table 6.3). Most water is present in enamel as free H₂O, the remainder in the form of OH groups within the crystals. Enamel mineral, commonly known as hydroxyapatite, has the general composition Ca₁₀(PO₄)₆(OH)₂ but consists of magnesium whitlockite, Ca₉Mg(HPO₄)(PO₄)₆, an apatite phase, Ca_{8.5}Na_{1.5}(PO₄)_{4.5}(CO₃)_{1.5}, and a slightly carbonated hydroxyapatite phase, Ca₁₀(PO₄)₆(OH,V,CO₃,F)₂. The magnesium content of enamel increases from the surface toward the interior; carbonate and magnesium are lost preferentially in slightly acidic solution.

Enamel has a prismatic structure with acicular (needle-shaped) 260 × 680 Å hydroxyapatite crystals, a surface area of 4 ± 1 m² g⁻¹ and a pore volume of about 9% (Figure 6.1). Mineral content is relatively constant but density varies from <2.86 g cm⁻³ at the dentin–enamel junction (DEJ) to 3.01 g cm⁻³ at the tooth surface. The carbonate content changes from <2% at the surface to >4% at the DEJ.

Enamel is permeable to water, ions, and small molecules. Under normal physiological conditions, enamel is in physicochemical equilibrium with saliva (i.e., stable). Nucleation of calcium phosphates is retarded/inhibited by salivary proteins, preventing continuous growth of the mineral phase of enamel. Being acellular, enamel does not undergo repair or replacement but undergoes change with age (Box 6.1).

Due to its high mineral content, enamel exhibits acid solubility and brittle behavior, fracturing easily if the underlying dentin is weakened by caries or undermined by improper cavity preparation.

6.2 Dentin

Dentin, comprising the tooth bulk, has a higher organic content than enamel, collagen comprising 85% of the organic portion. Dentin mineral [hydroxyapatite, Ca₈(PO₄)₄(CO₃)₂·5H₂O] has a higher Ca/P ratio than enamel and different levels of carbonate, magnesium, and sodium. Dentin crystals are platelike, 500–600 Å long with a slightly smaller width and a thickness of 20–35 Å.

Dentin matrix contains many proteins (Box 6.2) and surrounds tubules filled with odontoblastic processes in a serumlike fluid. The tubule characteristics vary with location (Table 6.4), probably due to greater crowding toward the pulp.

In the absence of odontoblastic processes, the tubules continuously contact the extracellular fluid of the pulp. Pulpal circulation maintains an intercellular hydraulic pressure of ca. 24 mm Hg, causing outward tubular fluid flow from the pulp toward the DEJ when enamel is removed. Similar flow also results from external hydrostatic and osmotic pressures. Positive or negative fluid flow through exposed tubules affects odontoblasts or pulpal nerve endings; this is the basis of the hydrodynamic theory of pulpal hypersensitivity (**hyperalgesia**).

Mechanical preparation of dentin forms a 3–15 μm mat of organic and inorganic particles, the **smear layer**, which partly/completely occludes dentinal tubules (Figure 6.2). The smear layer reduces hydrostatic pressures but has less effect on diffusion processes and, if removed by acid etching, chemical and bacterial products can diffuse toward the pulp against the pressure gradient.

6.3 Cementum

Cementum a hard connective tissue, similar to bone, covers tooth roots and provides for attachment of periodontal ligament fibers. It is about 50% mineralized with hydroxyapatite, the organic matrix being largely collagen and ground substance. Cementum is not vascularized and does not remodel but is more resistant to resorption than bone. The cementum layer is 20–50 μm at the cementum–enamel junction (CEJ), increasing to 150–200 μm toward the tooth apex.

Although cementum can be variable, two types are recognized:

- 1 Acellular:** Thin layer immediately adjacent to dentin surface of root
- 2 Cellular:** Covering the apical third of the root and overlying acellular cementum

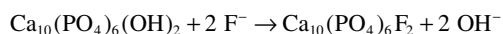
Cementum is formed by cementoblasts lining the root surface and deposition continues in phases throughout the life of the tooth. As acellular cementum forms, the cementoblasts retreat and deposit the cementum matrix.

6.4 Fluoridation of enamel

Fluoride incorporation decreases enamel acid solubility and increases caries resistance. Fluoride ingested from potable water or dietary sources forms fluorhydroxyapatite [Ca₁₀(PO₄)₆(OH,F)₂ or Ca₁₀(PO₄)₆F₂]. The optimum fluoride level in potable water is 1 ppm. Higher levels interfere with **amelogenesis**; fluoride levels >5 ppm interfere with **ameloblast** function, causing formation of mottled enamel.

Topical application of NaF to enamel forms CaF₂, a fast process accelerated by high fluoride content and low pH levels. Treatment depth extends to about 10 μm.

Topical agents with lower fluoride concentrations (e.g., APF gel) form fluorhydroxyapatite,

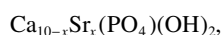


The reaction rate is slower, requiring longer contact times, but fluoride uptake is greater and the reaction profile is deeper (≤100 μm). Enamel reactivity with topical reagents decreases with higher natural fluoride contents.

The negative logarithms of the solubility products of hydroxyapatite (OHA) and fluorhydroxyapatite (FA) are $K_{\text{OHA}} = 117.2$ and $K_{\text{FA}} = 121.2$, so that fluoridation markedly decreases enamel solubility, particularly in low-pH and low-fluoride media.

6.5 Strontium

The **DMFT** score decreases linearly with increase in the strontium level in drinking water to about 10 ppm but the effect disappears at Sr levels >35 ppm. Strontium's effects are mostly in the pre-eruptive phase through its incorporation in deeper layers of enamel, particularly in the caries-susceptible magnesium whitlockite [Ca₉Mg(HPO₄)(PO₄)₆] phase. It appears that a solid solution of calcium hydroxyapatite and strontium apatite is formed with the following composition:



where 0 ≤ x ≤ 10.

Table 7.1 Composition of bone

Inorganic constituents (67 wt.%)	Organic matrix (33 wt.%)
Ca-deficient hydroxyapatite	25% Type I collagen
	5% Noncollagenous proteins:
	osteonectin
	osteocalcin
	bone morphogenic protein
	bone proteoglycan
	bone sialoprotein

Table 7.2 Inorganic constituents of bone (wt.%)

Constituent	Average content (%)	Range (%)
Calcium	35	33–36.5
Phosphorus	15	14.5–16
Carbonate	5	3–8
Sodium	0.45	0.26–0.59
Magnesium	0.45	0.32–0.78

Table 7.3 Average composition of bone mineral

Component	Presence (wt.%)
Magnesium whitlockite, $\text{Ca}_9\text{Mg}(\text{HPO}_4)(\text{PO}_4)_6$	20
Sodium- and carbonate-containing apatite, $\text{Ca}_{5.5}\text{Na}_{1.5}(\text{PO}_4)_{4.5}(\text{CO}_3)_{1.5}$	15
Defective hydroxyapatite, $\text{Ca}_9(\text{HPO}_4)(\text{PO}_4)_5(\text{OH})$	65

Table 7.5 Classification of bone types

Type of bone	Characteristics
Type I	Homogeneous, compact bone
Type II	Core of dense trabecular bone surrounded by a thick layer of compact bone
Type III	Thin layer of cortical bone surrounding a core of dense trabecular bone
Type IV	Core of low-density trabecular bone of poor strength encased in thin cortical bone

Table 7.4 Cells responsible for formation, maintenance, and resorption of bone

Cell	Type	Function
Osteoblasts	Uninucleated cells	Secrete bone matrix. Synthesize collagenous and noncollagenous bone protein (the osteoid). Mineralize osteoid.
Osteocytes	Entrapped osteoblasts in bone matrix	Osteocyte–osteoblast complex prevents bone hypermineralization.
Osteoclasts	Multinucleated cells	Remove mineral, after osteoblasts remove osteoid, by extracellular secretion of HCl and proteolytic enzymes that create an acidic environment that dissolves mineral and digests organic matrix.

The properties of mandibular and maxillary bone are central to dental implant success. Bone is a specialized connective tissue consisting of an organic matrix permeated by a poorly crystallized calcium-deficient hydroxyapatite. The mineral composition of bone differs from that of other dental hard tissues and bone has a greater organic content (Table 7.1 and Table 7.2).

Bone composition, approximating $\text{Ca}_7\text{Na}_2(\text{PO}_4)_3(\text{CO}_3)_3(\text{OH})$, varies from sample to sample. The mineral composition is given in Table 7.3.

The defective hydroxyapatite phase appears to dissolve preferentially at neutral pH values. The pH of bone fluid is slightly lower than that of the interstitial extracellular fluid of the noncalcified connective tissue of the body because of the slow exchange of calcium and phosphate from bone.

Based on gross appearance, bones are classified as long, short, irregular, sesamoid, or flat but all have the same inner structure:

- 1 Dense outer sheet of compact bone
 - (a) Perimeter is surrounded by osteogenic connective tissue membrane (periosteum).
 - (b) Internal surface of compact bone is covered by a single layer of bone cells separating the bone and marrow (endosteum).
- 2 Central medullary canal filled with red or yellow bone marrow
 - (a) Marrow cavity is interrupted along its length by a reticular network of trabecular (cancellous or spongy) bone.
 - (b) Entire surface of cancellous bone is covered by endosteum.
 - (c) Internal trabeculae support the outer, thicker cortical crust of compact bone.

Both compact and trabecular mature bone are composed of microscopic layers (lamellae), organized in three types of layering:

- (a) Circumferential lamellae (enclosing the entire bone and forming outer and inner perimeters)
- (b) Concentric lamellae (which constitute the bulk of compact bone and are the basic metabolic unit of compact bone, the osteon)
- (c) Interstitial lamellae (interspersed, and filling the spaces, between adjacent concentric lamellae)

Separate cells in bone (osteoblasts, osteocytes, and osteoclasts) are responsible for formation, maintenance, and resorption (Table 7.4).

Bone development occurs by three methods:

- (a) **Endochondral bone formation:** Occurs upon a cartilage matrix model
 - The cartilage model is resorbed as it is replaced by bone.
 - Also refers to the cartilage development immediately preceding bone.
- (b) **Intramembranous bone formation:** Occurs within a connective tissue membrane
- (c) **Sutural bone formation:** Occurs along sutural margins; a special case of intramembranous bone formation

With aging, bone may exhibit **osteoporosis** (localized mineral loss). Cellular activity decreases with age and results in a decrease in the

mucopolysaccharide proteoglycan-to-collagen ratio. Lowering of this ratio decreases the water content of bone and its permeability to ionic diffusion. Consequently, regions remote from the bone surface acquire a lower pH and there is local dissolution of bone mineral.

7.1 Dental implants and bone

Bone quality, both volume and density, determines implant success and has been classified into four types (Table 7.5). Greater implant success is found with Types I and II bone.

Bone density alone and the combination of bone volume and density are significant in implant success. Low bone volume combined with poor bone quality, i.e. Types III and IV bone, increases the prevalence of implant failure. Implant failures are more common in maxillae, where the bone is less dense, and with implants placed in severely resorbed mandibles. The mean failure rate with implant supported overdentures is 19% in the maxilla compared with 4% in the mandible.

Changes occur in bone, collagen, and bone proteins with advancing age, and fracture healing is longer in older patients. Thus, longer periods of healing after implant placement and before loading may be necessary for older patients.

Postmenopausal women can develop osteoporosis or osteopenia although postmenopausal estrogen status may only be relevant to implant success in the maxilla. Recent studies, however, indicate that osteoporosis drugs such as bisphosphonates, regardless of oral or intravenous administration, increase jaw necrosis. Even short-term use of osteoporosis drugs may leave the jaw vulnerable to necrosis.

Necrosis and reduced implant osseointegration can result if bone is heated above 47°C during implant site preparation because of collagen denaturation and necrosis of bone cells. A corollary to thermal damage is interfacial formation of connective tissue between implant and bone, leading to reduced integration and implant loosening.

Titanium, the optimal implant material, is inert and biocompatible. It cannot initiate new bone and blood vessel growth around the implant, which may limit implant osseointegration. Coating the implant with synthetic bone material (hydroxyapatite and/or bioglass) improves implant osseointegration.

Bone grafting can affect implant osseointegration, particularly if augmentation is required due to bone loss from periodontal disease, infection, or osteoporosis. Grafted bone must have time to integrate and mature to an organized structure since immature bone cannot withstand the torque inherent with dental implants while its replacement lamellar bone takes 6–12 months to evolve. The more organized structure of lamellar bone provides greater implant-to-bone contact and a better prognosis. Success rates for implants placed in grafted bone range from 77% to 85% compared with >95% success in mature, ungrafted bone.

Part II

Laboratory materials

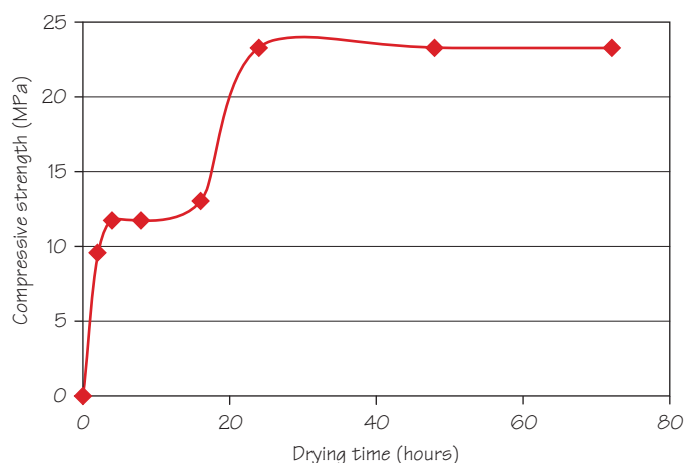


Figure 8.1 Effect of drying time on compressive strength of dental plaster.

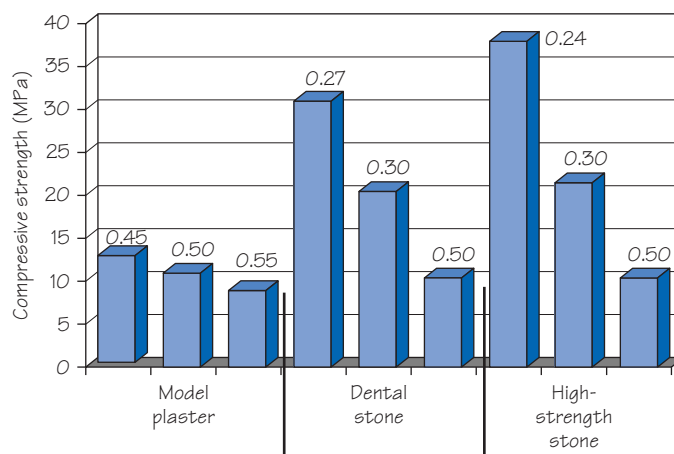


Figure 8.2 Effect of the W/P ratio (shown above the columns) on the strength of gypsum materials.

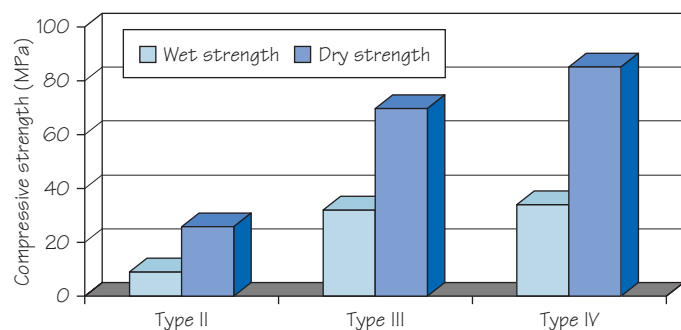


Figure 8.3 Wet and dry compressive strengths of gypsum materials.

Table 8.1 Dental gypsum products

ANSI/ADA specification nomenclature	Traditional nomenclature
Type I—Plaster, impression	Impression plaster
Type II—Plaster, model	Model or laboratory plaster
Type III—Dental stone	Class I stone; model stone
Type IV—Dental stone, high strength	Class II stone; die stone
Low expansion (ISO Type 4)	
High expansion (ISO Type 5)	

Table 8.2 Water-to-powder (W/P) ratios for gypsum materials

Product	W/P ratio (g water/100 g powder)
Plaster	40–50
Dental stone	25–30
Die stone	19–24

Table 8.3 Effect of W/P ratio on characteristics of mixed gypsum materials

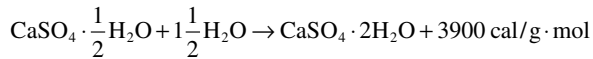
W/P ratio	Mix stiffness	Setting rate	Entrapped air	Pourability	Strength
Lower	Stiffer	Faster	More present	Reduced fluidity	Increased
Higher	Thinner	Slower	Reduced	Greater fluidity	Lowered

Table 8.4 Comparative properties of dental gypsum products

	Type II	Type III	Type IV
W/P ratio	0.45	0.28	0.24
Setting time (minutes)	12.0	8.0	7.0
Setting expansion (%)	0.30	0.18	0.10

8.1 Dental gypsum materials

Widely used in dentistry, gypsum materials are obtained from natural deposits of **gypsum**, $\text{CaSO}_4 \cdot 2\text{H}_2\text{O}$, which when heated loses 1.5 g mol of water and converts to the **hemihydrate**, $\text{CaSO}_4 \cdot \frac{1}{2}\text{H}_2\text{O}$. On mixing with water, the hemihydrate exothermically converts back to the **dihydrate**:



There are four types of “dental” gypsum (Table 8.1), which are chemically identical but differ in morphology and physical properties:

1 Model plaster: So-called beta-hemihydrate, produced by heating gypsum in an open kettle at 110–120°C; irregularly shaped particles that are porous and white

2 Dental stone: So-called alpha-hemihydrate, produced by dehydration of gypsum in water vapor under pressure at 125°C; yellow, with more uniformly shaped and denser particles than plaster

3 Die stone (high-strength stone): The highest density dental gypsum

- Type IV die stone is made by boiling gypsum in 30% CaCl_2 solution, washing out residual chlorides with 100°C water, and grinding the mass to powder.

4 Die stone (high strength, high expansion): New ultrastrength, high-expansion Type V die stone is discussed in Chapter 9

8.2 Setting reaction

Setting is due to the different solubilities of the di- and hemihydrates; during rehydration, a “dissolution center” surrounds the hemihydrate while a “precipitation center” forms around the dihydrate. The CaSO_4 concentration is higher in the dissolution center and lower in the precipitation center, where the less soluble $\text{CaSO}_4 \cdot 2\text{H}_2\text{O}$ precipitates out.

8.3 Water-to-powder (W/P) ratio

$\text{CaSO}_4 \cdot \frac{1}{2}\text{H}_2\text{O}$ theoretically requires 1.5 g mol H_2O per 1 g mol plaster (18.61 g water per 100 g plaster) but more water actually must be added; the three hemihydrates require different W/P ratios (Table 8.2). The W/P ratio affects the properties of all gypsum/water mixes (Table 8.3).

8.4 Factors in setting

8.4.1 Expansion

Theoretically $\text{CaSO}_4 \cdot \frac{1}{2}\text{H}_2\text{O}$ contracts by ca. 7% upon rehydration but actually expands 0.2–0.4% due to nucleation and outward growth of gypsum crystals from the supersaturated solution, causing simultaneous expansion and volumetric contraction. The materials are porous when set.

8.4.2 Spatulation

Spatulation speed and duration affect setting time and expansion because spatulation disrupts “precipitation centers,” forming new nuclei and reducing the setting time.

8.4.3 Water temperature

Water temperature affects (a) the relative solubilities of hemihydrates and dihydrates and (b) ionic mobilities. The hemihydrates and dihydrates have a 4.5:1 solubility ratio at RT, which drops to 1:1 at 100°C. The reaction rate drops with lower solubility ratios but Ca^{2+} and SO_4^{2-} mobilities increase with temperature rise, accelerating reaction rates so that setting times decrease up to 37°C. Reaction rates drop

and setting times lengthen at higher water temperatures until there is no reaction at 100°C.

8.4.4 Humidity

During manufacture not all $\text{CaSO}_4 \cdot 2\text{H}_2\text{O}$ converts to $\text{CaSO}_4 \cdot \frac{1}{2}\text{H}_2\text{O}$ and residual gypsum dehydrates to anhydrous CaSO_4 . The latter is hygroscopic, absorbing atmospheric moisture to form $\text{CaSO}_4 \cdot 2\text{H}_2\text{O}$ on the hemihydrate particle surfaces, affecting reactivity.

8.4.5 Colloids

Colloids adsorb onto the $\text{CaSO}_4 \cdot \frac{1}{2}\text{H}_2\text{O}$ surface and retard setting, resulting in a soft, easily abraded surface—an important effect when alginate impressions are poured up. Accelerators such as potassium sulfate (K_2SO_4) improve surface quality and reduce this problem.

8.5 Physical properties

8.5.1 Mixing

For the same W/P ratio, mechanical mixing decreases setting time, increases strength, decreases viscosity, and reduces expansion.

8.5.2 Setting time

Too rapid a set prevents proper fill of impressions during “pour-up” due to rising viscosity during setting but modifiers adjust the latter. Potassium sulfate and terra alba ($\text{CaSO}_4 \cdot 2\text{H}_2\text{O}$) accelerate setting. NaCl accelerates setting but increases setting expansion. Borax ($\text{Na}_2\text{B}_4\text{O}_7$) retards the setting reaction and a mixture of 0.1% CaO and 1.0% gum arabic reduces the water requirement of the mix.

8.5.3 Compressive strength

Compressive strength is proportional to dryness (Figure 8.1). Set gypsum requires at least 24 h and usually 7 d to lose excess H_2O and achieve adequate strength. Porosity in the set mass decreases strength. The strength of the set material is determined by the mix W/P (Figure 8.2).

8.5.4 Tensile strength

Gypsum is brittle and the one-hour diametral tensile strength (DTS) is only 50% of the dry strength. The ratio of compressive strength to DTS is 5–10:1.

8.5.5 Surface hardness

Surface hardness increases upon drying. Impregnation with resin monomers and subsequent polymerization or admixture of hardening solution containing colloidal SiO_2 increases hardness. Surface treatment with wax, oils, or glycerol improves carvability but not hardness.

8.5.6 Setting expansion

Gypsums expand on setting, with W/P ratios and additives ($\text{K}_2\text{SO}_4 \downarrow$, $\text{NaCl} \uparrow$) affecting expansion; setting of gypsum materials under water (**hygroscopic effect**) can increase expansion up to 100%. After setting, there is zero dimensional change with time.

8.6 Comparative properties

The W/P ratio, setting time, and setting expansion decrease while strength increases from Type II to Type IV products (Table 8.4 and Figure 8.3).

Dental stone and die stone are used as processing casts because of their greater strength, better abrasion resistance, and superior ability to record detail than plaster.

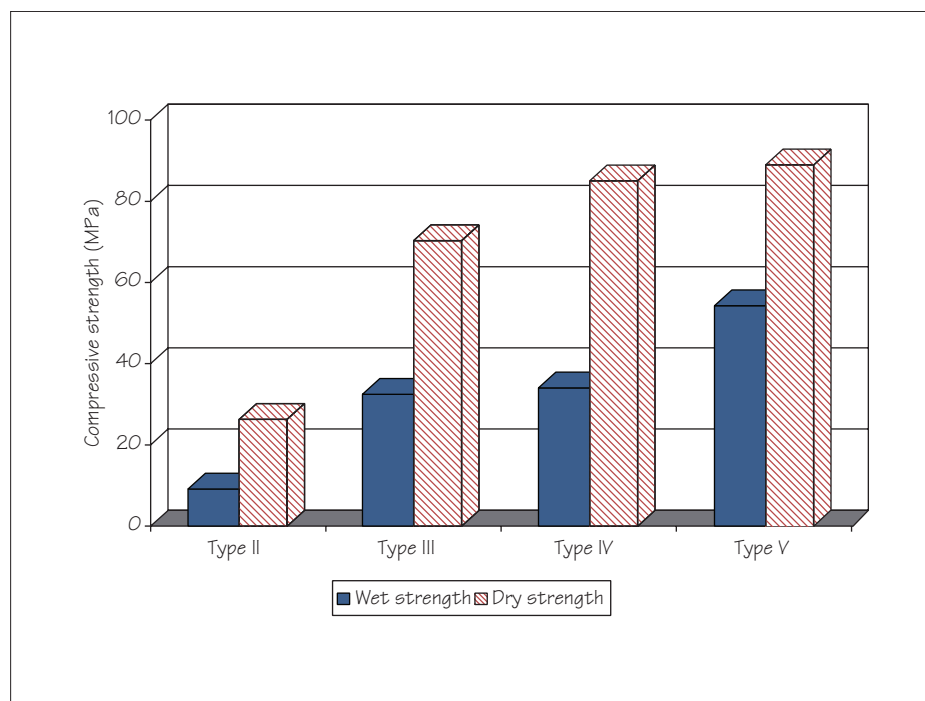


Figure 9.1 Compressive strengths of dental and die stones.



Figure 9.2 A cast prepared for construction of a crown. (Courtesy of Dentsply International.)

Table 9.1 W/P ratios, setting times, and setting expansions of dental and die stones

	Type II	Type III	Type IV	Type V
W/P ratio*	45	28	24	20–22
Setting time (minutes)	12.0	8.0	7.0	8–12
Setting expansion (%)	0.30	0.18	0.10	0.08

*W/P ratio in milliliters of H₂O per 100 g powder.

9.1 Gypsum products

The most widely used die materials are based on gypsum (Chapter 8), although only Types II, III, and IV are used for models. These hemihydrates are chemically identical but differ in morphology and physical properties.

1 Type II: Model plaster (so-called β -CaSO₄·½H₂O) has irregularly shaped and porous powder particles and is used to mount models.

2 Type III: Dental stone (so-called α -CaSO₄·½H₂O) has more uniformly shaped and greater density powder particles than plaster. Type III stone is used to pour study casts that are not being used to fabricate fixed restorations.

3 Type IV: Die stone (high-strength stone, also referred to as α -CaSO₄·½H₂O) has the highest density powder particles. Die stone is used to fabricate high-strength and abrasion-resistant dies used in fabricating fixed restorations.

4 Type V: The more recently introduced ultrahard, high-expansion die stone, manufactured by autoclaving gypsum, has greater strength than Types III and IV die stones and has optimal expansion for dies and for crown and bridge work. It is particularly suited for pouring polyvinyl and polyether impressions since there is less risk of model fracture during separation from the rigid elastomeric materials. It can also be used as an investing medium for casting gold alloys.

Dental and die stones have inherently greater strength than plaster due to lower water requirements and different powder morphologies (Table 9.1 and Figure 9.1).

9.2 Handling of gypsum materials

Theoretically CaSO₄·½H₂O should contract by ca. 7% on hydration but there is always a net expansion during setting (Table 9.1; also see Chapter 8). The properties of gypsum materials are modified by additives that adjust setting rate, setting expansion, and strength.

Compressive strengths of gypsum materials are proportional to dryness and inversely proportional to the W/P ratio (Figure 9.1). Accordingly, all gypsum materials must dry at least 24 h and usually 7 d to lose excess water and achieve maximum strength. Gypsum materials are brittle and have tensile strength one-fifth to one-tenth of compressive strength.

Surface hardness increases with evaporation of surface and subsurface water, and can be increased by impregnation with methylmethacrylate monomer or liquid (uncured) epoxy resin and subsequent polymerization; admixing with a hardening solution containing colloidal SiO₂ is also performed. Hardening treatment has little effect on abrasion resistance. Surface treatment with wax, oils, or glycerol improves carvability but has no effect on hardness.

Commercial die stones now are available in a variety of colors and there have been progressive improvements in the properties and handling characteristics of these materials due to changes in manufactur-

ing methodology and additives, notably resin enhancement. Modern gypsum-based die materials have enhanced properties including thixotropic behavior, which promotes smooth flow and permits stacking.

9.3 Polymeric die materials

There is a growing trend toward using resin die materials, which are more expensive but have greater versatility than gypsum materials. Many resin-based materials contain additives that permit digital scanning and often have an opaque, highly reflective color. Consequently, if these materials are used to create a restoration by the indirect technique with a light-cured material, the transmitted light reflects off the occlusal surface and the preparation walls, enhancing polymerization of the restorative resin from all directions. At least one variety of die material has been compounded with a dull, nonreflective surface, enabling its use for making silicone models for extraoral scanning during CAD-CAM procedures (see Chapter 42), which is useful when intraoral image capture cannot be performed.

Most new polymeric die materials are addition-curing silicones but are also referred to as polyvinyl siloxane (PVS) and vinyl polysiloxane (VPS) as well as A-Silicone resins. They are supplied in automix cartridges and, having high fluidity, can be injected directly into polyether impression materials from customized dispensing guns or an impression material delivery system. The advantages of these materials include excellent flow and ability to record minute detail. They set rapidly (within 2 minutes on the bench and faster in the mouth) to a rigid mass (90+ on the Shore Durometer D scale) but retain a degree of flexibility.

Polymeric die materials are smooth-surfaced, are extremely accurate, and exhibit minimal if any dimensional change. Further, PVS dies are extremely durable and will not crack, abrade, or chip if dropped; they can be trimmed with a scalpel blade without dust generation. A major advantage of these materials is that as soon as the dies have been poured, the model base can be created with a rapid-setting mousse.

9.4 Pouring the impression

Because of their higher strength and abrasion resistance, Type IV and V die stones are used for pouring elastomeric impressions to produce models for making the final restoration. Maximizing the properties of the die material and minimizing air incorporation and voids is accomplished by mixing the stone with a motorized vacuum mixer that simultaneously extracts air and spatulates the mix at a constant rate.

When pouring up the impression, the tray should be on a vibrating platform, which will facilitate flow of the thixotropic stone–water mix. After setting, impression and die stone should be gently separated slightly on one side and then the other, this incremental approach minimizing the risk of material breaking off from the model. A cast prepared for construction of a crown is shown in Figure 9.2.

Table 10.1 Dental waxes

Wax type	Application
Pattern	
Baseplate wax	Establishing initial arch during denture making
Casting (modeling) wax	Wax rims and temporary bases during denture making
Inlay wax	Making direct or indirect patterns for cast restorations
Processing	
Boxing wax	Perimeter border of impression trays during pour-up
Sticky wax	Temporary adhesive
Utility	Variety of applications
Impression	
Bite wax	Recording occlusal and jaw relationships
Corrective (impression) wax	Dental impressions

Table 10.4 Wax melting ranges and thermal expansions

Wax	Melting range (°C)	Approximate temperature range (°C)	Coefficient of thermal expansion (10 ⁻⁶ /°C)
Paraffin	40–71	20–28 28–34	307 1631
Microcrystalline	60–91		
Barnsdahl	70–74	22–40 40–52	185 243
Ozokerite	Ca. 65		
Montan	72–92	22–42 42–52	188 294
Carnauba	84–91	22–52	156
Ouricury	79–84	22–43 43–52	186 307
Candelilla	68–75	22–40 40–52	182 365
Japan wax	Ca. 51	22–39 39–45	304 755
Beeswax	63–70	22–41 41–50	344 1048
Inlay wax (hard)		22–38 38–45 45–50	323 629 328

Table 10.2 Natural waxes

Mineral	Plant	Animal
Paraffin	Carnauba	Beeswax
Microcrystalline	Ouricury	Spermaceti
Barnsdahl	Candelilla	
Ozokerite	Japan wax	
Ceresin	Cocoa butter	
Montan		

Table 10.3 Effect of wax additions on the properties of paraffin wax

Admixed wax	Effect on paraffin wax
Microcrystalline	Reduced volumetric change on solidification
Barnsdahl	Increased melting range, greater hardness, reduced flow
Ozokerite	Improved properties in melting range
Ceresin	Greater hardness, increased melting range
Montan	Increased hardness and melting range
Carnauba	Increased hardness and melting range, modified flow
Ouricury	Increased hardness and melting range
Candelilla	Increased hardness
Japan wax	Greater tackiness and emulsifying ability
Cocoa butter	Greater tackiness and emulsifying ability
Beeswax	Variety of effects to improve properties

Dental waxes are **thermoplastics** that are solid at RT, melt when heated, and harden without decomposition on cooling. Three categories are recognized, described in Table 10.1.

10.1 Composition

Dental waxes are classified by composition (Table 10.2); most are paraffins with other waxes, gums, oils, and resins added to modify properties (Table 10.3).

10.1.1 Mineral waxes

- **Paraffin waxes:** Paraffins are straight-chain alkanes with 26–30 carbon atoms; melting range increases with molecular weight (MW) and is decreased by oils ($\leq 0.5\%$ oil). On solidification and cooling, paraffins volumetrically contract 11–15% nonuniformly down to RT because of numerous phase transitions.

- **Microcrystalline waxes:** These are branched-chain hydrocarbons (41–50 C atoms) of greater MW and melting range than paraffins; they are tougher, are more flexible, and exhibit lower volumetric contraction. Affinity for oils facilitates hardness and tackiness modification.

- **Barnsdahl:** Barnsdahl wax is used to increase melting range and hardness while reducing flow of paraffin waxes.

- **Ozokerite:** A straight- and branched-chain hydrocarbon microcrystalline earth wax, ozokerite has a high oil affinity; 5–15% additions to paraffin waxes improve physical properties in the melting range of 54°C.

- **Ceresin:** Ceresin is a straight- and branched-chain hydrocarbon distillation product with higher MW and greater hardness than paraffin waxes; additions raise paraffin wax melting range.
- **Montan waxes:** Montan waxes are a mixture of long-chain esters and high MW alcohols, acids, and resins extracted from lignite, with similar properties to plant waxes. They are hard and brittle, blend well with other waxes, and raise the melting range and hardness of paraffin waxes.

10.1.2 Plant waxes

- **Carnauba and ouricury waxes:** These waxes are mixtures of straight-chain esters, alcohols, acids, and hydrocarbons, have high hardness, and are brittle with high melt temperatures. Additions of 10% can increase the melting range of paraffin waxes by 24° as well as increase hardness, but additions above 10% have no effect.
- **Candelilla waxes:** Candelillas comprise 40–60% paraffin hydrocarbons (29–33 C atoms) with free alcohols, acids, esters, and lactones. They harden paraffin waxes with little effect on melting range.
- **Japan wax:** A fat containing glycerides of palmitic, stearic, and higher MW acids, Japan wax is tough, malleable, and sticky; it increases the tackiness and emulsifying ability of paraffin wax.
- **Cocoa butter:** A fat composed of glycerides of palmitic, stearic, oleic, lauric, and lower MW fatty acids, cocoa butter is brittle at RT and is used to reduce dehydration of soft tissues.

10.1.3 Animal waxes

- **Beeswax:** An insect wax that is a complex mix of esters, hydrocarbons, and high MW organic acids, beeswax is brittle at RT but plastic at body temperature.
- **Spermaceti:** Obtained from the sperm whale, spermaceti is composed mainly of esters; it was formerly used to coat dental floss but is little used now.

10.1.4 Synthetic waxes

Synthetic waxes include polyethylene, polyoxyethylene glycol, halogenated hydrocarbon, hydrogenated waxes, and wax esters. Polyethylene waxes have a MW of 2000–4000 while polyoxyethylene waxes are polymers of ethylene glycols with similar melting temperatures and hardnesses to natural waxes but are poorly compatible with other waxes. They are used to plasticize and toughen wax films.

10.1.5 Gums, fats, and resins

- **Gums:** Notably **gum arabic** and **tragacanth**, gums are viscous plant exudates with complex compositions; they harden in air and form sticky, viscous liquids with water.
- **Fats:** Fatty acid esters, tasteless, odorless, and colorless when pure, increase the melting range and hardness of compounded waxes.

Oils modify wax properties; e.g., hydrocarbon oils soften waxes while silicone oils improve polishability.

- **Natural resins:** **Rosin** (colophony), **copal**, **kauri**, and **mastic** are tree/plant exudates. **Shellac** is produced by insects. They are relatively insoluble in water and are used to harden natural waxes. Solutions in organic solvent carriers are film-formers, e.g. copal resin cavity varnish.
- **Synthetic resins:** The synthetic resins, which include polyethylene, polystyrene, and vinyl resins, are added to paraffin waxes to improve toughness, film-forming characteristics, and melting range.

10.2 Wax properties

10.2.1 Melting and thermal expansion

Waxes have melting ranges—not a single melting temperature—and large thermal expansions/contractions, which can be modified but not eliminated by compounding.

Expansion coefficients vary with temperature (Table 10.3) and are greater for mineral than plant waxes due to weaker secondary valence forces. Higher secondary forces of plant waxes are due to their high ester contents.

Phase transitions cause waxes to have ≥ 2 rates of thermal expansion (Table 10.4). Postcarving temperature changes of inlay waxes can affect casting accuracy.

10.2.2 Mechanical properties

Wax strengths and elastic moduli are low, are temperature dependent, and decrease with temperature rise. Elastic modulus of paraffin wax decreases 91% from 24°C to 30°C and for carnauba wax decreases 58% from 23°C to 37°C.

10.2.3 Flow

Flow depends on temperature and the magnitude and duration of load application; flow increases near the melting range, more so for mineral than plant waxes.

10.3 Dental waxes

10.3.1 Pattern waxes

Thermal effects distort pattern waxes, particularly on standing unrestrained, distortion increasing with temperature and time. Uniformly heating wax, carvers, and die before incremental wax application and refrigerating the pattern minimize thermal distortion and stress relaxation.

10.3.2 Baseplate wax

Baseplate waxes typically comprise 75–80% paraffin or ceresin with additions of beeswax, carnauba wax, and microcrystalline waxes or resins. They show minimal flow at RT but 90% flow at 37°C.

10.3.3 Casting waxes

Casting waxes exhibit flow behavior similar to inlay waxes, with a maximum flow of 10% at 35°C and at least 60% flow at 38°C. Casting waxes are ductile and must bend over double without fracture at RT.

10.3.4 Inlay waxes

Typically, inlay waxes contain 60% paraffin, 25% carnauba, 10% ceresin, and 5% beeswax, with flow adjusted by the carnauba content, higher melting range paraffins, and/or $\leq 1\%$ resin. **Type I inlay wax** for indirect patterns is soft with greater flow below and above oral temperature and with less thermal contraction than **Type II inlay wax**, which is used for the direct technique.

In the past (and confusingly!), Type I waxes were designated as the hard inlay waxes while Type II waxes were the soft inlay waxes.

10.3.5 Sticky wax

Sticky wax is composed of beeswax and rosin and is sticky when melted but hardens to a tack-free, brittle material at RT.

10.3.6 Impression waxes

Impression waxes have high flow and ductility; they cannot be used for undercuts due to their inability to deform elastically but may be used with elastic impression materials.

11 Investments and casting

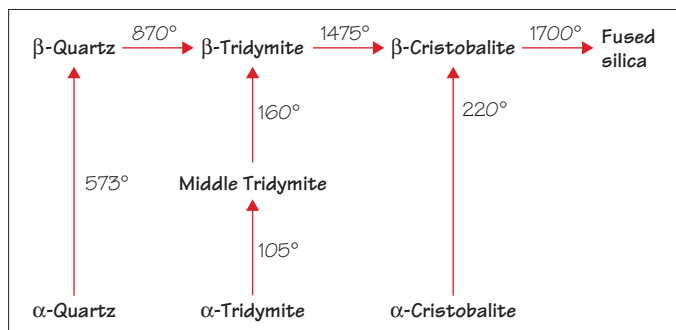


Figure 11.1 Thermal transformations of silica.

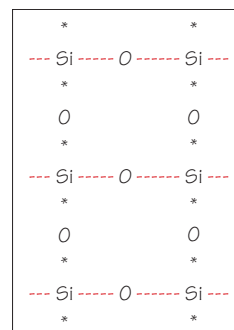


Figure 11.2 Polymerized silica.

Box 11.1 Requirements of investment materials

Easy manipulation
 Room-temperature (RT) and elevated temperature strength
 Rapid hardening
 High-temperature (HT) stability
 Thermal expansion over range of temperatures
 Sufficient porosity for gas escape
 Smooth surface finish
 Easy separation from the casting
 Nonreacting with the cast metal
 Moderate-to-low cost

Table 11.1 Expansion requirements for gypsum-bonded investments

Type	Application and expansion mode	Setting expansion (%)		Thermal expansion (%)	Combined expansion (%)
		Air	Water		
Type I	Inlay, thermal	0–0.6	—	1.0–1.6	1.3–2.2
Type II	Inlay, hygroscopic	—	1.2–2.2	0–0.6 (500°)	1.3–2.7
Type II	Partial denture, thermal	0–0.4	—	1.0–1.5 (700°)	1.2–1.9

Table 11.2 Thermal expansions of crystallographic forms of SiO_2

Crystallographic form	Expansion (%)	Temperature ($^{\circ}\text{C}$)
Quartz	1.4	600
Cristobalite	1.6	400
Tridymite	<1	600

Table 11.4 Effect of casting variables on porosity

Type of porosity	Sprue diameter increase	Sprue length increase	Raised melt temperature	Raised mold temperature
Localized	Decreased	Increased	Decreased	Decreased
Subsurface	Increased	Decreased	Increased	Increased
Microporosity	No effect	No effect	Decreased	Decreased

Table 11.3 Effect of manipulation variables on investment expansion

Factor	Setting and hygroscopic expansion	Thermal expansion
Increase in W/P ratio	Decreased	Decreased
Increased spatulation time	Increased	No effect
Increased spatulation rate	Increased	No effect
Increased age of investment	Decreased	No effect
Delayed immersion	Decreased	—
Higher water bath temperature	Increased	—
Wax strength	More distortion	Less effect
Sprue location	More critical	Less critical

Crowns, bridgework, inlays, and onlays are cast from metals by the **lost wax process**, in which a wax pattern is surrounded by **investment material** that hardens in position, forming a ceramic mold. The wax is then burned out to create the mold cavity.

11.1 Casting and investments

11.1.1 Composition

Investment materials must satisfy certain criteria (Box 11.1) and contain three types of material: **refractory**, **binder**, and **modifiers**. The refractory is one or more forms of **silica** (**quartz**, **tridymite**, or **cristobalite**) held together by a binder: gypsum, phosphate, or silicate. Gypsum-bonded investments are used for gold casting but palladium and base metal alloys require higher temperature binders, commonly phosphates. Modifiers (NaCl, H_3BO_3 , K_2SO_4 , graphite, Cu powder, or MgO) change physical properties.

11.2 Gypsum-bonded investment

11.2.1 Composition

Composed of 30–35% $\alpha\text{-CaSO}_4 \cdot \frac{1}{2}\text{H}_2\text{O}$, 60–65% quartz, and ca. 5% modifiers, gypsum-bonded investments set through addition of water by converting hemihydrate to gypsum, which binds the mass together. Silica provides strengthening during wax burn-out (when dihydrate reverts to hemihydrate) and thermal expansion. Upper temperature limit is 700°C, above which CaSO_4 breaks down, releasing SO_2 + SO_3 , causing embrittlement of the casting. NaCl and H_3BO_3 enhance thermal expansion.

11.2.2 Properties

Specified expansions for Types I, II, and III gypsum-bonded investments are indicated in Table 11.1. Minimum 2-h compressive strength for Types I and II is 2.41 MPa and for Type III, 4.82 MPa. Average setting time is 7–15 min, accelerated by prolonged and/or vigorous mixing but decreased by increased W/P ratio.

11.2.3 Effect of temperature

The three polymorphic forms of silica expand nonlinearly on heating due to transformations of the α -form (RT-stable) to β -form (HT-stable) but to different degrees (Table 11.2). Only α -forms of silica are used in investments. The α – β transformations (Figure 11.1), compensate for metal casting shrinkage.

During mixing, some water hydrates $\text{CaSO}_4 \cdot \frac{1}{2}\text{H}_2\text{O}$, the rest distributes in the setting mass and evaporates during investment heating. Some expansion occurs on heating to about 105°C after which the investment remains unchanged or contracts slightly up to 200° before expanding again, expansion varying with composition. On cooling, the investment exhibits an overall contraction. Investment reheating causes cracking.

11.2.4 Setting and hygroscopic expansion

All gypsum-bonded investments undergo **setting expansion (SE)**, **thermal expansion (TE)**, and **hygroscopic expansion (HE)**. SE is normally about 0.3% but increases to 1.3% with hygroscopic expansion due to investment contact with water during setting (e.g., placing casting ring in water bath, using a wet liner, or pouring water on investment surface). Expansions are affected by several factors, summarized in Table 11.3.

11.2.5 Other factors in expansion

1 Binder material: $\alpha\text{-CaSO}_4 \cdot \frac{1}{2}\text{H}_2\text{O}$ (stone) produces more expansion than $\beta\text{-CaSO}_4 \cdot \frac{1}{2}\text{H}_2\text{O}$ (plaster).

2 Particle size: Finer refractory (SiO_2) particles produce more expansion but $\text{CaSO}_4 \cdot \frac{1}{2}\text{H}_2\text{O}$ particle size has little effect.

3 Silica-to-binder ratio: Higher silica-to-binder ratios increase HE but decrease investment strength. Modifiers compensate for gypsum contraction below 700° and reduce the need for additional silica. Boric acid also compensates for gypsum contraction and strengthens the investment but disintegrates on heating, causing casting surface roughness.

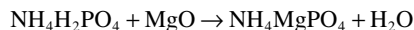
11.3 High-temperature investments

11.3.1 Phosphate-bonded investments

Phosphate-bonded investments contain three components: a source of PO_4^{3-} ions; a compound that reacts with PO_4^{3-} ions at RT; and a ceramic that hardens at high temperature. The binder content is usually <20%; the remainder is the refractory. Phosphate-bonded investments can be used for casting almost all metals. Two types are known: Type 1 for cast fixed restorations and Type 2 for removable restorations.

1 Monoammonium dihydrogen phosphate (MAP): MAP provides RT strength to investment and reacts with SiO_2 to increase high-temperature strength; dissolving in water provides PO_4^{3-} ions. Reaction with SiO_2 probably involves P_2O_5 in forming a **silicophosphate**. More MAP is present in formulation than is required for stoichiometry to ensure sufficient remains for reacting with SiO_2 .

2 Magnesium oxide: MgO reacts at RT with PO_4^{3-} ions, providing “green” strength:



3 Water: Water lowers the viscosity of the mix. Mixing investment with silica sol-based liquid provides higher SE and permits HE (absent with water mixes) while strengthening the investment.

11.3.2 Silica-bonded investments

Silica-bonded investments, now little used, comprise powdered quartz or cristobalite bonded with silica gel, which converts to silica on heating. Binder is **ethyl silicate**, hydrolyzing in the presence of HCl to **silicic acid sol** and ethanol:



The silicic acid sol reacts with quartz/cristobalite forming a silica gel polymer (Figure 11.2), providing the investment with greater SiO_2 content. Due to ethanol by-product, sodium silicate and colloidal silica are preferred binders to ethyl silicate. Investments have low porosity.

11.4 Considerations in casting

1 Metal shrinkage: The shrinkage is 1.25–1.7%, depending upon the metal and the casting shape and size; wax properties compensate for metal shrinkage.

2 Sprue diameter: The applied casting machine pressure and molten metal density determine metal flow rate into the mold cavity; faster flow occurs with larger sprues, higher pressures, and denser metals (Table 11.4). Flaring the sprue attachment to the bulkier portion of the casting and away from margins minimizes distortion.

3 Wax pattern: Positioning ca. 6 mm from end of casting ring optimizes amount of investment for strength versus thickness through which gases vent while promoting cooling of the casting.

Part III

Dental biomaterials

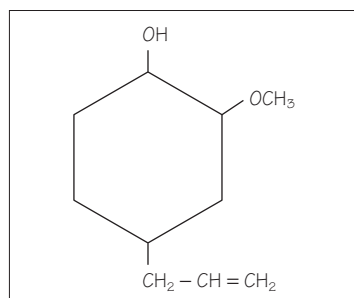


Figure 12.1 Structure of eugenol.

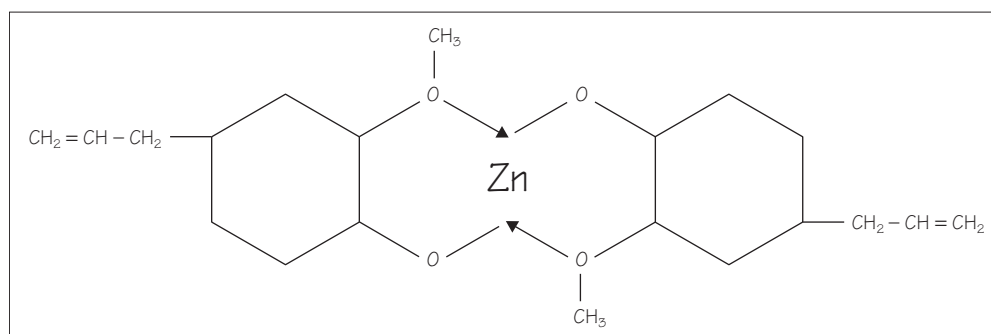


Figure 12.2 Structure of zinc eugenolate.

Table 12.1 Impression compound

Component	Addition level (%)
Thermoplastic (rosin, copal resin, carnauba wax)	47
Lubricant (stearic acid)	3
Filler (talc)	50

Table 12.2 Flow properties of impression and tray compound

Material	Flow at 37°C	Flow at 45°C
Type I (impression compound)	<6%	>85%
Type II (tray compound)	<2%	70–85%

Table 12.3 Zinc oxide–eugenol impression material

Base paste	Reactant
Zinc oxide, 85%	Eugenol, 15%
Inert oil, 15%	Rosin, gum, oil, 65%
	Filler (talc, kaolin), 16%

Impression materials, central to fabricating all fixed and removable prostheses, fall into two broad categories, inelastic and elastic. The former are used for edentulous patients and for interocclusal records whereas the latter (Chapter 13) are for dentate patients.

12.1 Factors in impression-making

An impression material must wet the hard and soft tissues and flow into crevices to accurately record detail, behavior determined by the material's flow properties.

With Newtonian fluids, flow is proportional to applied load but there is no such proportionality for non-Newtonian fluids; many impression materials are non-Newtonian. The yield value of a fluid is the minimum applied load for flow to occur whereas viscosity is its resistance to flow. **Thixotropy** is the term applied to a fluid or semirigid gel that resists flow when static but which flows upon agitation.

Most impression materials set by chemical reaction upon mixing, setting being accelerated by the temperature rise upon tray insertion into the mouth. During setting, both yield value and viscosity increase, with detail being lost if tray seating is delayed. Detail is also lost if tray movement occurs after seating due to disruption of the setting reaction.

12.2 Inelastic impression materials

Inelastic impression materials exhibit minimal elastic deformation under flexure and will fracture without plastic deformation when subjected to stresses greater than their tensile, compressive, or shear strength. They cannot be used for undercuts.

The principal inelastic impression materials are **impression plaster**, **impression compound**, and **zinc oxide–eugenol (ZOE)** paste.

12.2.1 Impression plaster

Predominantly plaster of Paris ($\text{CaSO}_4 \cdot \frac{1}{2}\text{H}_2\text{O}$), impression plaster contains additives to control setting time and expansion. The W/P ratio (approximately 0.60, higher than for dental plaster) affects mix consistency and the working and setting times. The setting reaction is the same as for plaster but proceeds faster.

Impression plaster is very accurate and has minimal setting expansion (<0.06%) due to antiexpansion agents such as K_2SO_4 that also accelerate setting.

When first mixed, impression plaster is highly fluid and requires special trays. It is **mucostatic** and can record fine details, often being used as a wash in a preliminary compound impression (see Section 12.2.2). It is useful for patients with flabby ridges, i.e. mobile soft tissues overlying the alveolar bone.

Impression plaster powder must be kept dry since water sorbed at high humidities extends the setting time.

12.2.2 Impression compound

Type I impression compound, or **modeling plastic**, is used for impressing the edentulous ridge as well as for single tooth impressions when contained in a copper cylinder or in impression trays for recording the entire arch. Primarily a wax mixture (Table 12.1), impression compound is supplied in sheets or rods that soften with heat. In use,

compound must be seated before it can cool. Having low thermal conductivity, impression compound must be warmed thoroughly in a water bath at 50°C for good flow. Flow increases rapidly on warming, from 2–3% at 34° to approximately 95% above 45°C.

Prolonged water bath immersion causes leaching of components; likewise, incautious heating with a flame can cause ignition and/or volatilization of constituents. Loss of low MW components causes compound to become grainy and brittle.

Type II impression compound is more viscous and flows less than Type I compound (Table 12.2); it is used to make primary impression trays. These custom trays are then provided with a thin layer or wash of a second impression material for a more detailed and accurate impression of the tissues. Impression compound also is used for border molding of stock and custom impression trays during fitting.

Impression compound is subject to stress relaxation, i.e. warping or distorting on standing, particularly at higher temperatures. Such changes cause inaccuracies but are minimized by allowing impressions to cool thoroughly after removal from the mouth and then pouring immediately. Separation of the poured cast from the compound tray is facilitated by softening in warm water.

12.2.3 Zinc oxide–eugenol

Zinc oxide–eugenol (ZOE) is a paste–paste system (Table 12.3) that sets in the presence of H_2O through interaction of zinc oxide and eugenol to form a **zinc eugenolate** matrix surrounding unreacted ZnO particles (Figure 12.1 and Figure 12.2). Some products use oil of cloves (containing 70–85% eugenol) rather than pure eugenol in the reactant (accelerator) paste to reduce the tissue burning sensation for patients.

ZOE impression pastes are classified as hard (Type I) and soft (Type II) pastes. Type I pastes set faster (ca. 10 minutes) and to a more rigid mass with greater strength than Type II pastes, which set in approximately 15 minutes. Setting rates are accelerated by adding a drop of water during mixing.

ZOE impression paste, mainly used for corrective/wash impressions, has excellent dimensional stability with a setting shrinkage of less than 0.1%. Unset ZOE is **mucostatic** and accurately records soft tissue detail. ZOE pastes tend to adhere to glass and mucosa.

Type II ZOE impression paste is used as a temporary denture liner.

12.2.4 Noneugenol pastes

Some products replace the eugenol with carboxylic acids, typically **orthoethoxybenzoic acid (EBA)**, to reduce the tissue burning sensation. These acids react with ZnO to form solid compounds similar to ZOE.

12.2.5 Surgical pastes

ZOE pastes may be placed over gingivectomy sites as protection against trauma and foodstuffs, and to promote healing. Surgical pastes have similar compositions to ZOE impression pastes but are softer and set more slowly. They have sufficient strength to resist shearing and fracture under masticatory forces.

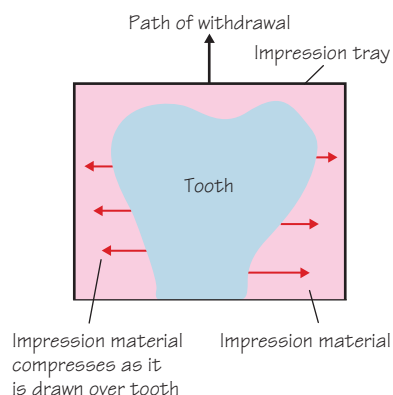


Figure 13.1 Compression of impression material at the bulbosity of the tooth as tray is withdrawn from the mouth.

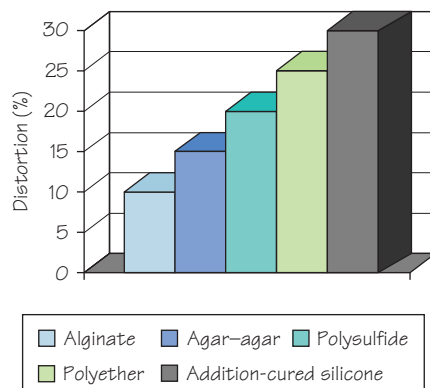


Figure 13.2 Approximate permissible distortions for 100% recovery of elastomerics.

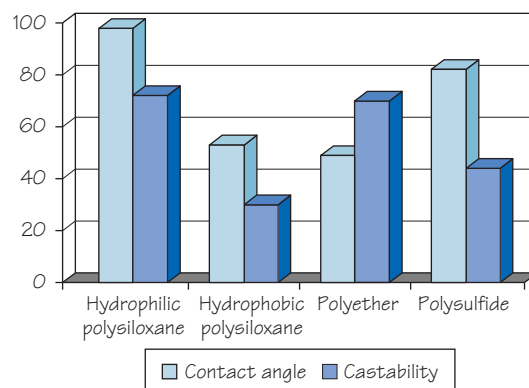


Figure 13.3 Contact angle of water and castability of dental gypsum into impression materials.

Table 13.1 Approximate composition of alginate impression material

Component	Amount (%)	Function
Sodium/potassium alginate	15	Water soluble, reacts with Ca^{2+}
Calcium sulfate	14	Reacts with potassium alginate to form insoluble gel
K_2SO_4 , silicates, borates	10	Counterbalance gel effect on gypsum
Sodium phosphate	2	Retarder for Ca^{2+} /alginate reaction
Diatomaceous earth	70	Filler
Colorant, flavoring, disinfectant	Minor/trace	

Table 13.2 Problems with alginate impressions

Grainy material	Irregular voids	Distortion
Improper mixing	Moisture/debris	Delayed pouring
Prolonged mixing	Poor mixing	Tray movement during set
Excessive gelation		Premature tray removal
Incorrect W/P ratio	Rough/chalky cast	Incorrect tray removal
	Impression not clean	Excessive gelation
Tearing	Impression too wet	
Inadequate bulk	Premature cast removal	Bubbles
Moisture contamination	Delayed cast removal	Overmixing of material
Premature tray removal	Incorrect mix of gypsum material	Poor mixing technique
Prolonged mixing		

Table 13.3 Polysulfide elastomeric impression material

Base paste	Catalyst
Polysulfide polymer, 80–85%	Lead dioxide, 60–68%
Filler (TiO_2 , ZnSO_4 , CuCO_3 , or silica), 16–18%	Dibutyl phthalate, 30–35%
	Sulfur, 3%
	Minor ingredients, 2%

Table 13.4 Setting behavior of principal medium-body elastomeric impression materials

Material	Viscosity at 45 s (cP)	Working time (min)	Setting time (min)	Dimensional change (% at 24 h)
Polysulfide	110,000	3–6	6–8	–0.45
Polyether	130,000	2–3	3–4.5	–0.24
Addition-cured silicone	150,000	2–4	4–6.5	–0.17

Table 13.5 Mechanical properties of medium-body elastomeric impression materials

Material	Permanent deformation (%)	Strain in compression (%)	Flow (%)	Shore A hardness	Tear strength (g/cm)
Polysulfide	3–5	11–15	0.5–1	30	3000–7000
Polyether	1–2	2–3	0.02	35–60	2800–4800
Addition-cured silicone	0.05–0.3	2–5	0.01–0.03	50	2200–3500

Elastic impression materials are required for dentate patients because as they are drawn over the bulbosity of the tooth, they compress against the tray (Figure 13.1), and elastic, more compressible materials have greater latitude in use. Approximate distortion (compression) values for 100% recovery are shown in Figure 13.2. After tray removal from the mouth, the impression requires about 30 minutes to recover before pouring up.

13.1 Alginate (irreversible) hydrocolloids

Alginate, a powder/water (P/W) system setting by sol–gel transition, is widely used for primary impression taking and for study casts. It has good elastic properties but poor long-term stability. The powder, containing potassium alginate (soluble salt of anhydro- β -D-mannuronic acid), calcium sulfate, and other components, is mixed with water to form a gel (Table 13.1).

The setting reaction is $\text{K-alginate} + \text{CaSO}_4 \rightarrow \text{Ca-alginate} + \text{K}_2\text{SO}_4$, retarded by Na_3PO_4 , which preferentially reacts with Ca^{2+} until exhausted; formation of insoluble Ca alginate causes mix gelation; additional agents provide desirable properties.

Alginates are hand-mixed and require careful proportioning since the P/W ratio and mixing affect the setting reaction, gel strength, consistency, and flow. Alginates are thixotropic and must be vigorously mixed, placed in the tray, and seated quickly to ensure accuracy. Poorly mixed material is grainy.

Gels are weak and have very low tear strength such that alginate cannot be used in thin section. Tray movement or premature tray removal causes inaccuracies, and distortion and tearing are reduced if the tray is removed rapidly from the mouth. Permanent set is determined by % compression, time under compression, and time of recovery.

Gels are subject to **syneresis** (loss of moisture and components) and/or **imbibition** (sorption of fluids); pour-up cannot be delayed or the impression left in contact with gypsum for extended periods. Most problems are caused by incorrect proportioning, poor mixing, or poor technique (Table 13.2).

13.2 Agar-agar (reversible) hydrocolloid

Agar-agar is a sol at elevated temperature and forms a viscoelastic gel on cooling; it requires a special conditioning unit and water-cooled trays for impression-taking. It is similar in properties to alginates but agar has better elastic recovery. Agar-agar is little used now.

13.3 Polysulfide rubber

This material has a polysulfide (mercaptan) polymer base (Table 13.3) that reacts with sulfur and is catalyzed by PbO_2 to form a high MW cross-linked rubber. Initial set occurs in 8–10 min with setting shrinkage but reaction continues for several hours, leading to long-term dimensional change. About 50% of setting shrinkage occurs within the first hour so that pour-up should be prompt. It has good elastic properties and excellent tear strength but an unpleasant taste and is messy in use. It is available in different consistencies or “**bodies**.” Other catalyst systems, e.g. $\text{Cu}(\text{OH})_2$, are less effective than PbO_2 .

13.4 Condensation-cured silicones

Base material is a dimethyl siloxane with 35–75% silica or CaCO_3 filler; the catalyst is stannous octoate and alkyl silicate. Polymerization liberates alcohol with high setting shrinkage. These silicones are rapid setting, somewhat rigid, and now little used.

13.5 Addition-cured silicones (polysiloxanes)

The base is a filled low MW polymer with silane groups that react with catalyst paste containing low MW polymer with vinyl groups and chlorplatinic acid catalyst. Polysiloxanes addition-cure with little dimensional change; they have long-term stability and are hydrophobic, but hydrophilic addition-cured silicones incorporate surfactants. They have reasonable tear strength but are somewhat rigid (high modulus), with minimal flow; they are available in different consistencies and have excellent elastic recovery (i.e., minimal permanent deformation).

13.6 Polyether impression materials

The base is a polyether with terminal imine rings that undergo cross-linkage catalyzed by 2,5-dichlorobenzene sulfonate; the thinner is octyl phthalate. Use is variable; they are rapid setting, with moderate dimensional change on setting, good elastic recovery, and high modulus; they can induce allergic reactions; and they are susceptible to moisture absorption.

13.7 Properties of elastic impression materials

Modern impression materials are supplied in automixing dispensers, and mechanical mixers with dynamic-mixing tips and with base and catalyst pastes stored within the unit are now available.

13.7.1 Setting properties

Heavier body materials have high viscosities. Viscosity increase upon setting is faster for polyethers and silicones, i.e. shorter working times and faster sets, than for polysulfides. Addition-cured silicones have the lowest polymerization shrinkage whereas light-body condensation-cured silicones have the highest (Table 13.4).

13.7.2 Mechanical properties

High hardness and large **strain in compression** (high modulus) indicates greater force requirement to remove impression from mouth. Polysulfides have the greatest flexibility (lowest modulus); see Table 13.5. Addition-cured silicones have the least permanent deformation (i.e., greatest elastic recovery) and polysulfides have the most.

Silicones and polyethers have lower flow and greater hardness than polysulfides. Polysulfides have the greatest tear strength, thin sections resisting tearing on removal from sulci or interproximal regions. All materials exhibit viscoelasticity, necessitating rapid removal from mouth.

Addition-cured hydrophilic silicones and polyethers have the highest wettability (Figure 13.3) and are most easily poured up.

13.8 Tray adhesives

Alginates require perforated or lock-rim trays but custom trays with adhesives are preferred for elastomeric materials. Polysulfide rubber adhesive is butyl rubber or styrene/acrylonitrile dissolved in chloroform or a ketone. Silicone adhesives are often a reactive silicone, e.g. poly(dimethylsiloxane), which bonds to the elastomer, and ethyl silicate, which forms a hydrated silica that bonds to the tray. Slight roughening of tray and adhesive application ahead of use improves retention.



Figure 14.1 Clinical photograph showing wax block-outs of missing maxillary posterior teeth and a lower RPD in place. (Courtesy of Dentsply International.)



Figure 14.2 Occlusal registration taken with Regisil® vinyl polysiloxane (VPS) bite registration material. (Courtesy of Dentsply International.)

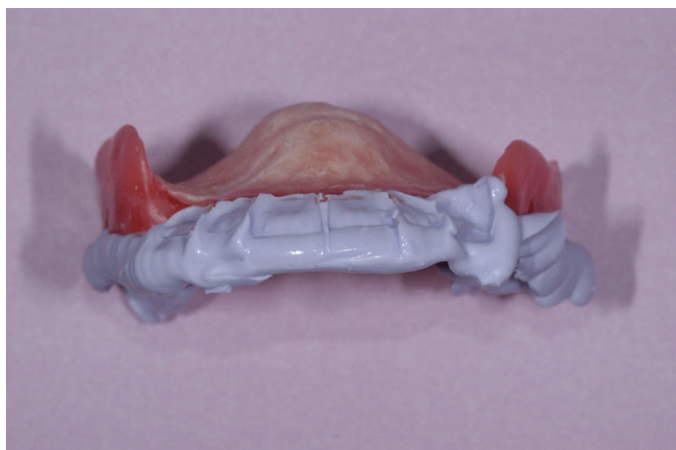


Figure 14.3 Wax maxillary denture base fabricated on VPS registration. (Courtesy of Dentsply International.)



Figure 14.4 Casts of maxilla and mandible with VPS occlusal registration in place. (Courtesy of Dentsply International.)

Box 14.1 Requirements of an occlusal registration material

- Ease of handling
- Good consistency
- Thixotropic behavior
- Ability to wet the dentition
- Fast set
- Flexibility
- No brittleness during trimming

When replacing teeth and hard tissue, occlusal—or bite—registration is necessary to record the relative positions of the upper and lower jaws and the dentition when the jaws are brought into apposition in centric relation. This ensures that the restored tooth will be correctly aligned and interdigitating with the opposing teeth on normal closing of the jaws.

For satisfactory occlusal registration, the recording medium must satisfy certain criteria (Box 14.1). Ease of handling is rarely a problem with modern materials since they are cartridge-delivered and have sufficient flow (i.e., consistency) to adequately and evenly spread over (i.e., wet) the teeth before starting to set.

After placement, the recording medium must remain in position without dislodgement by the patient's tongue. Satisfying this requirement depends in part on **thixotropy** and the setting rate; slower setting materials, e.g. mousse-type media (see Section 14.2), are more susceptible to dislodgement than their faster setting poly(vinyl siloxane) counterparts. Thixotropy in the registration medium allows it to be applied to the teeth without slumping or running while presenting minimal resistance to bringing the opposing dentition into occlusion.

Flexibility is important since the set registration material often must be withdrawn from undercuts and interproximal spaces without tearing, distortion, or fracture. Finally, trimming excess material from the registration medium should be possible without risk of fracture; a sharp scalpel is usually sufficient for this although some of the harder, more rigid materials require trimming with a rotary instrument.

14.1 Traditional registration materials

The simplest approach to occlusal registration uses softened wax in the patient's mouth and having the jaws close normally. This method is simple and rapid but lacks accuracy. Further, wax is subject to stress relaxation, undergoing distortion on heating or cooling.

Occlusal registration using more accurate and stable media such as zinc oxide–eugenol pastes was the norm for many years. Although ZOE is more accurate and dimensionally stable than wax, it is still a rigid material and will fracture on withdrawal from undercuts or approximal regions.

14.2 Modern registration materials

Addition-cured vinyl polysiloxane (VPS) mousse-type materials are now widely used occlusal registration media. Mousse materials are very versatile in that they do not slump, set rigidly, are dimensionally accurate, and have good elasticity with a Durometer-85. Available with different setting rates, they can be used without a tray. Although mousse materials are becoming standard for this procedure, there is increasing use of puttylike VPS materials because of their fast set (≤ 1 minute), high elasticity and tear resistance, excellent elastic recovery, and minimal post-setting and long-term dimensional change. As noted previously, most registration materials are supplied in automix

cartridge systems, although easy-mixing two-pack systems based on both VPS putty systems and polyether impression materials are also available. Examples of the clinical use of a commercial VPS material, Regisil®, are shown in Figure 14.1, Figure 14.2, Figure 14.3, and Figure 14.4. The advantages of VPS materials include a high tear strength so that they can be used with deep undercuts and to accurately record interproximal details.

At least one bis-acryl material is available in automixing cartridges. Bis-acryl materials tend to be slower setting and have a lower viscosity than VPS materials, so they are more difficult to dispense. Further, bis-acryl materials tend to be more rigid than their VPS counterparts and can lock into undercuts.

Most registration materials are relatively firm, with a Shore-A final hardness of 80–90, and snap set within one minute. Although a fast set is advantageous, it can be problematic when recording full arches. Further, occlusal registrations taken with very hard and rigid materials must be handled with a degree of care since they can break if dropped onto a hard surface.

Whereas most registration materials are colored and flavored, some products are transparent, which is useful in visually confirming total closure when taking an occlusal registration. However, this transparency can impede distinguishing between matrix and die material following pour-up and it is not unknown for dental laboratories to miss seeing transparent occlusal registrations when cases are sent to them.

14.2.1 Pattern resins

Pattern resin materials also can be used for occlusal registration, especially when there are multiple abutments. These are basically self-curing powder-and-liquid acrylic resin systems, one of the most popular being a quick-cure, low-shrinkage material primarily intended for dental laboratory use. These powder and liquid components are mixed to a dough that may be placed directly in the mouth or used in the indirect technique for casting patterns. It is advisable with the indirect technique to use a lubricant or separating medium to avoid migration of the monomer into the stone model.

The advantages of these materials are that they set quickly, are rigid, and exhibit little or no shrinkage; however, they do contain monomer that can cause allergic reactions in sensitive individuals. Further, there is a setting exotherm, and flushing with water when setting in the mouth is advisable to avoid soft-tissue damage. In use, a thin mix is made of the material, applied to the dentition, and covered with a sheet of tinfoil or cellophane. The teeth then are brought into occlusion until the material sets.

Although it is customary to use a tray with an occlusal registration material, it is possible to undertake this procedure without a tray, particularly with mousse materials, puttylike materials, and pattern resins. Many modern occlusal registration materials can be used as a scannable occlusal index for CAD-CAM systems (Chapter 42).

15 Precious metal alloys

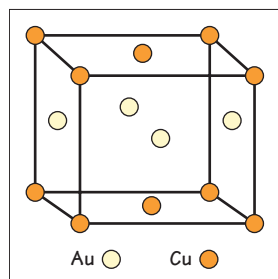


Figure 15.1 Face-centered tetragonal superlattice of AuCu.

Box 15.1 Requirements of casting alloys

Castability
Solderability
Accuracy of fit
Mechanical strength
Corrosion resistance

Box 15.2 Noble metals

Gold
Platinum
Palladium
Iridium
Rhodium
Osmium
Ruthenium

Table 15.1 Effect of hardening on gold alloys

	As-cast	Hardened
Proportional limit (MPa)	359	587
Tensile strength (MPa)	483	794
Elastic modulus (GPa)	97	100
Elongation (%)	15	5
Hardness (BHN)	130	220

Table 15.2 Dental gold casting alloys

Metal/alloy*	Characteristic	Application	Fusion temperature (°C)
Pure gold	Very soft	Direct fillings	1063
Type I [83]	Soft	Occlusal inlays	1005–1070
Type II [78]	Medium	Intracoronary inlays	900–970
Type III [78]	Hard	Onlays, ¾ crowns, crowns, bridgework	875–1000
Type IV [75]	Very hard	Removable partials	875–1000

*[. . .] denotes minimum noble metal content.

Table 15.3 Average properties of dental casting alloys

Alloy	Type I	Type II	Type III	Type IV
Au (%)	81–83	76–78	73–77	71–74
Pt (%)	—	—	—	0–1
Pd (%)	0.2–4.5	1–3	2–4	2–5
0.2% Yield strength (MPa)	105 [105]*	165 [165]	220 [300]	290 [495]
Tensile strength (MPa)	300 [300]	340 [380]	425 [530]	495 [770]
Elongation (%)	27 [27]	33 [33]	35 [17]	33 [6]
Vickers hardness (VHN)	65 [65]	97 [120]	135 [160]	160 [240]

*[. . .] denotes values in hardened condition.

Table 15.4 Average properties of low-gold casting alloys

	A	B	C
Noble metal (%)	52–65	35–45	22–30
Yield strength (MPa)	425 [810]*	425 [570]	220 [230]
Tensile strength (MPa)	570 [880]	570 [705]	425 [440]
Elongation (%)	17 [3]	12 [5]	17 [12]
VHN	185 [280]	185 [265]	150 [165]
Melting range (°C)	900–1025	1080–1150	1100–1150

*[. . .] denotes values in hardened condition.

Table 15.5 Casting conditions for low-gold alloys

Alloy	Melting range (°C)	Heat source	Investment
Type A	Ca. 1000	Gas/air torch	Gypsum-bonded
Types B and C	1000–1150	Gas/oxygen torch	Phosphate-bonded

Table 15.6 Heat treatments for gold alloys

Hardening	Softening
Heat at 450° for 2 minutes	Heat to 700°
Slow cool to 250° over 30 minutes	Water quench to room temperature
Water quench to room temperature or	
Heat at 350° for 10–15 minutes	
Water quench to room temperature	

Table 15.7 Casting defects

Problem	Possible cause
Discoloration	Breakdown of investment Incomplete wax elimination Contamination of gold Carbon inclusions
Distortion	Distortion of the wax pattern Type of pattern Type of wax Wax thickness
Incomplete casting	Insufficient mold venting Incomplete wax elimination Incorrect investment W/P ratio Insufficient casting pressure High metal surface tension High metal viscosity
Irregularities	Air bubbles Water film Incomplete wax elimination Investment heated too rapidly Investment damaged by impact of molten alloy
Porosity	Incomplete wax elimination Solidification shrinkage causing localized shrinkage porosity, microporosity, or subsurface porosity Gas effects causing pinhole porosity, gas inclusions, or subsurface porosity Entrapped air
Surface roughness	Breakdown of investment Overheating of alloy Excessive casting pressure Investment composition Incorrect investment W/P ratio Foreign body inclusions

Table 15.8 Effect of casting variables on porosity

Type of Porosity	Sprue diameter increase	Sprue length increase	Raised melt temperature	Raised mold temperature
Localized	Decreased	Increased	Decreased	Decreased
Subsurface	Increased	Decreased	Increased	Increased
Microporosity	No effect	No effect	Decreased	Decreased

Dental casting alloys must satisfy certain criteria (Box 15.1). Pure gold has excellent corrosion resistance but is soft, malleable, and ductile, requiring alloying for adequate strength.

15.1 Gold and noble metals

1 Pure gold: Now of limited use, gold formerly was used as a direct filling material because it welds to itself under pressure. Gold is embrittled by Pb, Bi, and Hg and typically is alloyed with Cu, Ag, and Pt to improve its mechanical properties.

2 Noble metals: The noble metals are elements with a high electropositive potential, i.e. resist electrochemical corrosion. They are dense, have a high melting point (MP), and resist oxidation at temperature (Box 15.2).

3 Platinum: Pt foil is used as a matrix for fused porcelain (PFM) restorations because of oxidation resistance, high MP, and an expansion coefficient similar to porcelain that prevents metal buckling or porcelain fracturing during temperature changes. Pt additions of $\leq 8\%$ to gold improve its hardness and elasticity but lighten its color.

4 Palladium: Pd has lower cost and is less dense than Pt and has the lowest MP of noble metals; it is used as alloying element for Au and Ag but whitens color of Au. White golds contain large amounts of Pd and Ag. Pd occludes hydrogen when heated, requiring care during melting.

5 Iridium, ruthenium, and rhodium: These are added in small amounts (ca. 50 ppm) as grain refiners for casting alloys.

15.2 Precious metals

Precious metals have high cost but are not necessarily noble (e.g., silver). Silver is a precious but non-noble metal that is ductile, malleable, highly conductive, and harder than Au but has a lower MP. It is susceptible to corrosion (improved by Pd addition), especially in S^{2-} and Cl^{-} media. Pure Ag occludes air and O_2 on melting, reduced by 5–10% additions of Cu. Ag readily alloys with Au and reduces red color of Au–Cu alloys.

15.3 Gold alloys

15.3.1 Carat and fineness of gold

Gold content is traditionally designated by the carat ($1k = 1/24$ of the gold content) or fineness; these terms now are rarely used in dentistry.

15.3.2 Gold–copper alloys

Au and Cu form a continuous series of solid solutions, stable above $424^{\circ}C$ but transforming into the ordered phases AuCu and $AuCu_3$ below $424^{\circ}C$. The face-centered cubic $AuCu_3$ phase structure is not found in dental alloys. AuCu has face-centered tetragonal structure, the unit cell being a cube with Au atoms at the center of the side faces (Figure 15.1). The high-temperature cubic lattice transforms to a tetragonal lattice at lower temperatures, hardening the alloy by inducing localized strains that inhibit dislocation movement.

Specification of gold content of yellow alloys is 62–92.5 wt.% or 34–78 wt.%. AuCu contains 75% gold by weight but only 50% of the number of atoms; also, increasing Cu content decreases MP. Slow cooling to RT allows transformation to ordered tetragonal AuCu with attendant strength increase (Table 15.1), which is the basis of gold alloy heat treatment. A minimum 75% gold content is required for corrosion resistance and for good castability.

15.3.3 Dental gold casting alloys

Casting alloy **Types I–IV** are ternary **Au–Ag–Cu alloys** with a fifth category used for PFM restorations (Chapter 17). Alloy Types I–IV melt over a temperature range that decreases with increasing alloying additions (Table 15.2). White golds with high Pt and/or Pd content have higher fusion temperatures (1100 – $1150^{\circ}C$) but high-Ag alloys are similar to yellow gold.

Complex restorations require higher strength alloys (Table 15.3), whereas softer metals are used for simple inlays. The average modulus of elasticity is 90 GPa, little changed by heat treatment. Additions of Ir or Ru improve mechanical properties.

15.3.4 Low-carat gold alloys

Low-carat alloys have gold content below ANSI/ADA specifications but their mechanical properties satisfy requirements. **Type A** alloys contain 40–60% Au and Pd. **Type B** alloys contain 10–20% Au with higher Pd content than **Type C**, which are Pd–Ag alloys with no Au; their properties (Table 15.4) are comparable to Type III and Type IV alloys but are less ductile. Order of corrosion resistance is Type A > Type B > Type C. They have higher fusion temperatures than high-gold alloys and require different casting conditions (Table 15.5).

15.4 Heat treatment of gold

Hardening or softening of gold castings depends upon temperature and conditions (Table 15.6), and the alloy must contain copper. Types I and II alloys are little affected by heat treatment. Note that prolonged heating at 700° causes grain growth and reduced strength.

15.5 Problems with castings

Causes of common problems with castings are indicated in Table 15.7 and the effects of casting variables on porosity are summarized in Table 15.8.

Table 16.1 Dental applications of base metal alloys

	Cast metals	Wrought metals
Nickel–chromium	Partial denture frameworks	—
	Crowns and bridges	—
	Metal–ceramic restorations	—
Cobalt–chromium	Partial denture frameworks	—
Cobalt–chromium–nickel	—	Orthodontic wires
Stainless steel	—	Orthodontic wires and brackets
	—	Endodontic instruments
Titanium and Ti alloys	Crowns and bridges	Implants
	Partial dentures	Crowns and bridges
Nickel–titanium	Implants	Orthodontic wires
	—	Orthodontic wires

Table 16.2 Compositions of cobalt–chromium and nickel–chromium casting alloys

	Cobalt–chromium	Nickel–chromium
Chromium	≥25%*	≥ 20%*
Molybdenum	≥4%	≥ 4%
Beryllium	Trace	≤ 2%
Cobalt	Main constituent	—
Nickel	0–30%	Main constituent
Chromium + cobalt + nickel	≥85%*	≥ 85%*

All alloys contain 0.2–0.5% carbon, 0.4–0.6% silicon.

*ADA/ANSI and ISO specification requirements

Box 16.1 Characteristics of titanium and titanium alloys

Benign biological response
 High corrosion resistance
 Rapid repassivation (10^{-9} seconds)
 Low density (4.5 g/cm^3)
 Low elastic modulus (100 GPa)
 High strength (240–550 MPa)

Table 16.3 Average properties of Co–Cr and Ni–Cr alloys

Property	Cobalt–chromium	Nickel–chromium
Density (g/cm^3)	8	8
Fusion temperature ($^{\circ}\text{C}$)	≤1500	≤1350
Casting shrinkage (%)	2.3	2.0
Tensile strength (MPa)	850	600
Proportional limit (MPa)	700	500
Modulus of elasticity (GPa)	220	185
Vickers hardness	400	350
Elongation (%)	1.5	2.5

The dental applications of base metal alloys are indicated in Table 16.1.

16.1 Cast chromium alloys

16.1.1 Cast chromium alloys

Cobalt–chromium and **nickel–chromium** alloys are being used increasingly for crowns and bridges and for **removable partial denture** (RPD) frameworks (Table 16.2). Ni–Cr alloys are comparable to Type III golds but are rarely used for all-metal cast restorations whereas Co–Cr alloys, comparable to Type IV golds, are used for RPDs. Properties of Co–Cr and Ni–Cr alloys are compared in Table 16.3.

16.1.2 Alloying elements

Chromium provides **corrosion resistance** but alloys with >30% Cr are difficult to cast and can form brittle σ phase. **Cobalt** and/or **nickel** increase strength, hardness, and modulus (Co has greater effect). **Carbon** content strongly affects properties: too much is embrittling but too little lowers strength, while it can form carbides with many elements. **Nitrogen** content must be low (<0.1%) to avoid embrittlement.

Addition of 3–6% **molybdenum** has a strengthening effect. **Aluminum** in Ni-alloys strengthens by formation of Ni_3Al whereas 1–2% **beryllium** lowers fusion range by about 100°C. **Silicon** and **manganese** increase **fluidity** and **castability**. Ni and Co are interchangeable in Co–Ni–Cr alloys but increasing Ni and decreasing Co with careful control of Mo and C produces alloys with higher strength and improved ductility.

16.2 Investments and casting

Phosphate- or silica-bonded investments are required but their lower porosity (compared with gypsum-bonded investments) may trap gases, causing casting voids. Casting can be technique-sensitive due to influence of mold and metal temperature and spruing on casting properties. Careful melting is required to avoid oxidation and carbide/nitride formation with prolonged heating or excessive temperatures. Casting shrinkage, $\geq 2\%$, requires greater mold expansion than for golds, achieved mainly by thermal expansion. Good extraction facilities are required for casting and grinding of Ni–Cr alloys due to beryllium and nickel in dust; allergic/sensitive patients may react to Ni-containing alloys.

16.3 Microstructure

Alloys have complex microstructures that vary with casting conditions. Yield and fatigue strength are controlled by effect of solute atoms (C, Cr, and Mo) on dislocation movement: reduced C levels increase ductility but decrease yield strength; excess grain boundary carbides decrease ductility but complete absence of carbides decreases strength.

16.4 Heat treatment

Heating can reduce strength and ductility, necessitating low temperatures and minimal time when soldering or welding of Co-base RPDs to avoid adverse effects.

16.5 Physical and mechanical properties

Base metal alloys have higher melting points than gold alloys (800–1050°C) with lower densities than low and high golds (12 and 18 g/cm³, respectively) so that castings are lighter than gold castings of the same dimensions. Base metals have greater elastic moduli and hardnesses but lower ductilities than hardened gold alloys, resulting in greater stiffness and decreased tolerance to bending. Mechanical polishing is difficult due to high hardness but polish is retained well in clinical service.

16.6 Titanium and titanium alloys

Titanium and Ti alloys are advantageous in dental applications (Box 16.1), notably because of biocompatibility, high strength, and corrosion resistance.

16.6.1 Pure titanium

Commercially pure (c.p.) titanium is available in four grades depending upon oxygen (0.18–0.40 wt.%) and iron (0.20–0.50 wt.%) content. Room-temperature lattice of Ti (α phase) transforms to β phase at 883°C; if a component is predominantly β phase, it is stronger but more brittle than α -phase Ti. α -Ti is weldable but difficult to work at RT; β -Ti is malleable at RT; ($\alpha+\beta$) alloys are strong, formable, but difficult to weld.

Industrial Ti casting has been known for >50 yr but aerospace and medical components commonly are produced by powder metallurgy (hot isostatic pressing, or **HIP**). Ti is not cast in the dental laboratory by conventional methods because of high MP (ca. 1700°C for c.p. Ti) and high reactivity (reacts with H₂, O₂, and N₂ above 600°C) whereas its low density makes centrifugal casting difficult. New techniques allow casting of Ti (and Ti–13Cu–4.5Ni) into crowns and partial and complete denture bases while alloying Ti with Pd and Cu lowers melting point. Mechanical properties of cast c.p. Ti are similar to Type III and IV golds but crowns can have porosity if casting conditions are not carefully controlled. Cast c.p. Ti complete denture bases are less satisfactory than crowns.

16.6.2 Titanium alloys

Alloying elements stabilize either the α or the β phase by affecting the ($\alpha+\beta$)-to- β transformation temperature.

- **Ti–6Al–4V** is a two-phase ($\alpha+\beta$) alloy at RT that transforms at 975°C to β alloy; its microstructure depends on whether working and heat treatment (HT) are performed above or below transition temperature by determining relative amounts of α and β . With this alloy, Al stabilizes the α phase by raising the transformation temperature whereas V, Cu, and Pd stabilize the β phase by lowering it. Microstructure determines mechanical properties of this alloy, which is widely used in implants.
- **Ti–6Al–4V** and **Ti–15V** have similar properties (except for elastic modulus) to Ni–Cr and Co–Cr alloys. Cast Ti–6Al–4V can be refined temporarily with hydrogen to improve physical properties.

Box 17.1 Requirements for PFM restoration materials

Inherent strength and rigidity
 Strong metal–ceramic bond
 Good esthetics
 Durability
 Soft-tissue compatibility
 Low thermal conductivity

Box 17.2 Requirements for metalloceramic alloys

Strength and rigidity
 High fusion temperature
 Nondeforming of porcelain
 No effect on porcelain color
 Strong bond to porcelain
 Similar coefficient of expansion to porcelain

Box 17.3 Problems with PFM restorations

Poor transverse strength
 Brittleness
 Need to carry porcelain over occlusal surface
 Wear of opposing dentition
 Expensive laboratory procedures
 Difficult to repair

Table 17.1 Average compositions of noble metal ceramic-bonding alloys

Element (wt.%)	1	2	3	4	5	6
	High gold	Low gold	Low gold–silver	Palladium–silver	Palladium–copper	Palladium–cobalt
Gold	85	48	51		0–2	
Palladium	6	43	28	53–88	76	88
Platinum	7					
Silver	1		15	35		
Copper					10–15	
Cobalt						4–5
Casting temperature (°C)	1150	1325	1325	1330	1180	1343

Additions of 0–2 wt.% of tin, indium, and iron are present in gold alloys.

Table 17.2 Average properties of noble metal ceramic-bonding alloys

	High gold	Low gold	Low gold–silver	Palladium–silver	Palladium–copper	Palladium–cobalt
Tensile strength (MPa)	490	720	670	640	690–1300	790
Elastic modulus (GPa)	95	116	103	100	95	122
Elongation (%)	4	10	10	12	9	25
Hardness (kg/mm ²)	175	220	220	220	375	220
Density (g/cm ³)	18.6	13.5	13.8	10.9	10.6	11.0

Table 17.3 Compositions of base metal alloys for metalloceramic restorations*

	Nickel-base	Cobalt-base	Titanium
Chromium	11–20	15–20	
Molybdenum	4–6	0–4	
Tungsten	0–2	0–4	
Aluminum	0–5	0–2	0–6
Beryllium	0–2		
Vanadium			0–4
Casting temperature (°C)	1300–1450	1350–1450	1760–1860

*Low additions of Fe, Ga, Ru, or Sn harden the alloys and provide bonding to porcelain.

Table 17.4 Properties of base metal metalloceramic alloys

	Nickel-base	Cobalt-base	Titanium
Tensile strength (MPa)	400–1000	520–820	240–900
Elastic modulus, (GPa)	150–210	145–220	103–114
Elongation (%)	8–20	6–15	10–20
Hardness (kg/mm ²)	210–380	330–465	125–350
Density (g/cm ³)	7.5–7.7	7.5–7.6	4.4–4.5

Metal–ceramic or porcelain-bonded-to-metal (PFM) restorations are popular because of their combination of strength and esthetics (Box 17.1), but the metal coping must satisfy many requirements (Box 17.2). Most notable is the need for the formation of a strong bond between metal and the ceramic mass without the metal coping melting or undergoing creep during porcelain firing. Bonding depends upon formation of oxides on the metal surface that permit bonding to the porcelain, but these metal oxides must not interact with the porcelain and cause staining or discoloration. Four types of alloy satisfy these criteria: high-gold alloys, low-gold alloys, palladium alloys, and chromium-containing alloys, the latter now the alloys of choice based on cost.

17.1 PFM restorations

PFMs comprise a cast metal coping onto which high-expansion porcelain is fired. Porcelain has a thermal expansion coefficient of $13.0\text{--}14.0 \times 10^{-6}/^{\circ}\text{C}$ compared with $13.5\text{--}14.5 \times 10^{-6}/^{\circ}\text{C}$ for most metals, and the slightly greater metal contraction after firing places the porcelain under compressive stress, reducing sensitivity to tensile forces. There are certain disadvantages to the PFM restoration (Box 17.3), and all-ceramic restorations (ACRs) are becoming increasingly popular (Chapter 41).

17.2 Ceramic-noble metal systems

Average compositions of noble metal alloys for PFM restorations are listed in Table 17.1 and their physical properties are indicated in Table 17.2.

1 High-gold alloys (Type 1): gold–palladium–platinum alloys have good corrosion resistance and are easily cast and soldered, but they have moderate strength and poorly resist thermal distortion. Platinum and palladium increase the melting range and decrease the thermal expansion coefficient while iron strengthens by forming FePt₃ precipitates through heat treatment (30 minutes at 550°C) during the porcelain firing cycle. No copper is present, to avoid staining of the ceramic. These alloys contain indium and/or tin, which oxidize more readily than gold and platinum, and facilitate bonding to the ceramic. Oxide formation occurs during degassing or pre-oxidizing before porcelain application.

2 Low-gold alloys (Types 2 and 3): Type 2 alloys have higher palladium content but no platinum or iron. Type 3 alloys contain silver, which can cause problems with porcelain. Indium, gallium, tin, or cobalt effect solid solution hardening, lower the fusion temperature, and improve bonding to porcelain. These alloys have good mechanical properties, low specific gravities, high melting temperatures, and good corrosion resistance and are readily cast and soldered.

3 Types 4, 5, and 6 alloys: These are palladium alloys that contain silver, copper, and/or cobalt for strengthening and for bonding to porcelain. Their properties improve during porcelain firing but they form dark oxides. Type 4 alloys contain more silver but otherwise are similar to Type 3 alloys. Type 5 alloys contain copper, have high strength and moderate ductility, and are easily cast and soldered but they poorly resist thermal distortion. The cobalt-containing (Type 6) alloys have properties similar to Type 4 alloys but are more ductile.

17.3 Ceramic-base metal systems

Three types of base metal alloys are used for PFM restorations; compositions and physical properties are summarized in Table 17.3 and Table 17.4. These alloys are harder and have higher fusion temperatures than noble metals but are more technique-sensitive and more difficult to cast and finish. Their cost is markedly lower than for precious metals.

1 Nickel-base alloys: These alloys contain 11–20% Cr for corrosion resistance and are **strengthened by aluminum or titanium** through formation of Ni₃Al or Ti₃Al. Beryllium additions facilitate porcelain bonding and lower the fusion temperature. Alloy properties are variable but strength and elastic modulus are greater than for noble metals.

2 Cobalt-base alloys: These alloys contain chromium (15–20%) for corrosion resistance and sometimes 3–6% ruthenium for improved castability. They strengthen by solid solution hardening. They are less variable in properties than nickel alloys and also are stronger than noble metals.

Although nickel- and cobalt-base alloys have higher elastic moduli than noble metals, coping thickness cannot be reduced because metal thickness is a major factor in the transverse strength of the coping–ceramic system.

3 Titanium alloys: Commercial purity (c.p.) titanium and Ti–6Al–4V alloy have excellent corrosion resistance and good physical properties. Titanium alloys are difficult to cast but they can be machined or spark-eroded to form copings. Ceramic bonding to titanium is possible but requires the use of special porcelains.

18 Implant metals



Figure 18.1a Modern dental implant. (Courtesy of American Dental Implant.)



Figure 18.1b Modern tapered dental implant. (Courtesy of BioHorizons Inc.)



Figure 18.2 Radiograph showing osseointegration of implants in bone.

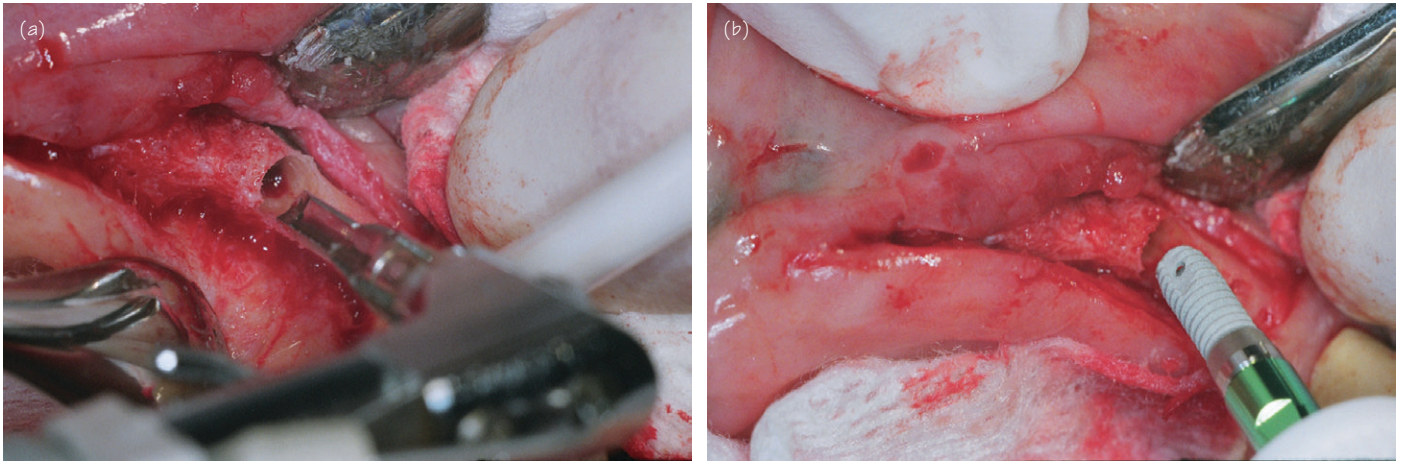


Figure 18.3 (a,b) Site preparation and placement of a “skinny” implant. (Courtesy of American Dental Implant.)

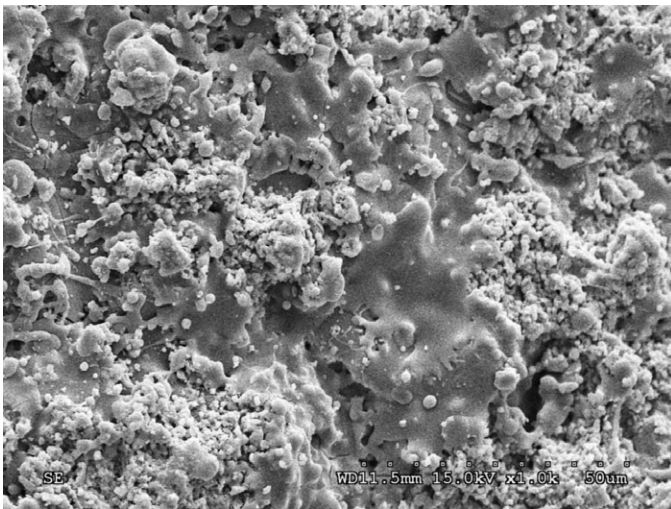


Figure 18.4 Hydroxyapatite (HA) coating on the surface of a dental implant (scanning electron micrograph, x1000). (Courtesy of BioHorizons Inc.)

Over the millennia, a variety of approaches were sought to replace missing teeth and there is clear archeological evidence that the Mayans were the first to use endosseous implants (implants embedded into bone) over 1400 years ago to support dental prostheses. Since then, notably in the second half of the 20th century, there were many attempts to alleviate the denture retention problems associated with resorbed ridges and other pathological conditions that prevented or severely restricted conventional prosthodontic therapy for the edentulous patient.

These approaches included the use of triangulated pins, commonly fabricated with the chemically inert tantalum, to support pontics. Another approach was to cast from cobalt–chromium alloys subperiosteal implants that closely conformed to the residual ridge and that were buried beneath the mucosa but had abutments projecting through the mucosa onto which were placed dentures. Finally, the Linkow blade vent systems, developed in the 1950s by Leonard Linkow, the “father” of modern implant dentistry, were used with great success for

many years. The Linkow implants comprised titanium (and other alloy) blade-shaped endosseous implants inserted into the jaw bone onto which were placed dentures or single units to replace missing dentition.

The modern “screw” implant (Figure 18.1a and Figure 18.1b) derives from the pioneering work of Stefano Tramonte in Italy and Per-Ingvar Brånemark in Sweden, who both advocated the use of titanium for dental implants. The excellent physical properties and outstanding biocompatibility of titanium (see Chapter 16) were the driving force for this application. In particular, Brånemark described the clinically observed close apposition and adherence of bone with titanium, which he termed **osseointegration** (Figure 18.2). Since then, a wide variety of “screw” or tooth-root-shaped endosseous implants have come into clinical use and they have achieved remarkable clinical success such that they are now considered important components of the restorative dentistry armamentarium. However, the clinical success of dental implants requires good clinical technique, accurate placement, and careful patient selection with good bone quality (see Chapter 7), Figure 18.3a,b.

18.1 Implant alloys

Of central importance for any metal within the oral cavity is corrosion resistance as well as mechanical strength. Consequently, the vast majority of modern dental implants are fabricated from titanium and its alloys, notably Ti–6Al–4V, the so-called 6–4 alloy, although c.p. titanium and alloys such as Ti–13Cu–4.5Ni also have been evaluated (see Chapter 16). It appears that most implants are fabricated using powder metallurgy, typically hot isostatic pressing (HIP) technology.

The efficacy and rate of osseointegration of bone and implant have been enhanced by techniques such as designing the implant with a screw profile, providing a microtexture to the implant surface, as well as coating the surface with hydroxyapatite (HA; Figure 18.4). More recently, a novel approach to dental implantology has been adopted in which the implant surface has been coated with a nanometer-thick layer of protein containing a bisphosphonate drug. Animal studies indicate that the bone surrounding the implant becomes denser and stronger, ensuring a more durable metal–bone interface than found with other approaches.

19 Partial denture base materials



Figure 19.1 Cast Vitallium removable partial denture. (Courtesy of Dentsply International.)



Figure 19.3 Corrosion and surface tarnishing of a chromium-cobalt removable partial denture framework due to the action of a hypochlorite-containing denture cleanser.

Table 19.1 Base metal casting alloys

	Jelenko LG	Nobilium	Vitallium	Ticonium
Chromium	27	30	30	15
Nickel	13	—	—	—
Molybdenum	4	5	5	5
Aluminum	—	—	—	5
Iron	1	—	1	0.5
Manganese	0.7	—	0.5	5
Other	—	—	—	<0.5% Be
Balance	Cobalt	Cobalt	Cobalt	Nickel

Note: All alloys contain 0.2–0.5% C, 0.4–0.6% Si.



Figure 19.2 Removable partial denture with a metal base mounted on a cast, showing the acrylic gumwork, prosthetic teeth, and the denture clasps positioned on the abutment tooth.



Figure 19.4 Flexible resin partial denture. (Lucitone FRS, courtesy of Dentsply International.)

Table 19.2 Mechanical properties of partial denture alloys

Alloy	Yield strength (MPa)	Tensile strength (MPa)	Ductility (% elongation)	Elastic Modulus (GPa)	Hardness (VHN, kg/mm ²)
Jelenko LG	495	675	10.0	228	300
Nobilium	565	825	1.6	228	380
Vitallium	495	640	1.5	228	380
Ticonium	690	800	1.7	186	340
Type IV gold (hardened)	480–510	750–790	5–7	90	220–250

Table 19.3 Physical properties of flexible partial denture materials

	Lucitone FRS	Valplast	Flexite	Acetal resin
Tensile strength (MPa)	62.1	37.9	68.9	68.9
Flexural strength (MPa)	68.9	40.0	89.6	96.5
Izod notched impact	7.4	0.9	1.4	2
Melting point (°C)	240–270	186	187	175
Glass transition temperature (°C)	155	34	—	162
Shrinkage (%)	0.7	—	—	—

A **removable partial denture (RPD)** is a dental prosthesis that replaces one or more teeth and the associated structures of the maxilla or mandible. The prosthesis comprises artificial teeth attached to a base that is supported through contact with the underlying oral tissues or implants and attached to abutment teeth with clasps or precision attachments. The denture base or framework may be fabricated from metal or a resin. Denture teeth are discussed in Chapter 20.

19.1 Metallic RPDs

Most dentists and prosthodontists favor the use of a metal alloy partial denture framework with **acrylic resin (PMMA)** “gumwork” and either resin or porcelain prosthetic teeth (Figure 19.1). The use of a metal framework ensures the rigidity of the RPD base, reducing the risk of damage from flexure of the denture to the gingival cuff surrounding the abutment teeth.

Metal partial denture frameworks are commonly cast from nickel-chromium or cobalt-chromium alloys and, increasingly rarely, from gold alloys. It is generally accepted that any metal used for a cast RPD framework should have a yield strength of at least 415 MPa in order to resist permanent deformation when used as clasps (Figure 19.2).

The composition and properties of precious metal and base metal dental casting alloys are discussed in Chapters 15 and 16. The compositions and properties of some widely used metals for RPDs are summarized in Table 19.1 and Table 19.2. All base metal alloys contain chromium for corrosion resistance and have superior strengths, elastic moduli, and Vickers hardness values than gold but, in most cases, are less ductile.

Because of their greater elastic moduli and hardnesses but lower ductilities than hardened gold alloys, base metals are stiffer and less tolerant of bending than gold. Further, their high hardness values make base metals more difficult to polish than gold alloys, but, once polished, they are better able to retain their surface finish in clinical service.

The clasps and rests are commonly cast directly with the framework, although clasps that are required to have special characteristics or are replacements for broken clasps may be soldered to the framework. Silver solders (high melting Ag-Cu-Zn alloys) are used for this purpose, but care must be taken with exposed solder joints to ensure that they do not corrode in immersion-type oxygenated denture cleaners (see Chapter 24). Corrosion is not generally a problem when soldered joints are “buried” within the acrylic gum work.

Since the framework and clasps are fabricated from nickel-chromium or cobalt-chromium alloys (and, far less commonly, gold alloys), corrosion rarely presents a problem with RPD frameworks within the oral cavity. There is, however, a risk of corrosion of the metal framework with immersion in bleaches and hypochlorite-containing cleansers (Figure 19.3).

The PMMA gumwork will stain with use, particularly when regularly exposed to highly colored beverages such as fruit juice, tea, and coffee as well as tobacco products. Regular cleansing is necessary to maintain an esthetic appearance as well as to eliminate mouth odor and remove bacterial accumulation; see Chapter 24.

19.2 Polymeric RPDs

Removable partial dentures fabricated from polymers such as PMMA and nylon are often supplied to patients until a cast metal RPD from

a dental laboratory is available. Polymeric RPDs may be provided to patients as final or long-term prostheses because of economic necessity but few prosthodontists advocate their long-term use because they tend to be bulky. Further, because polymeric RPDs tend to flex in use, they may cause damage to the periodontal interface between the soft tissues and abutment teeth, ultimately causing separation. Another consideration is that resin-based RPDs have low thermal conductivity and do not permit thermal stimulation of underlying tissues. Polymeric RPDs, however, have appeal to patients because of cost considerations.

The light-cured resins, based on urethane oligomers, have superior mechanical properties compared with acrylic resins. Clinical data indicate that there are no indications of allergic reactions or tissue irritation, fit is improved, and there is greater stain resistance compared with conventional acrylic resins.

Many patients dislike the rigidity of acrylic and metallic framework RPDs, and allergic reactions to poly(methyl methacrylate) are not uncommon. Accordingly, there has been a growing interest in the clinical use of more flexible denture base materials that are fabricated wholly or in part from other resin systems, e.g. **polyamides** such as **nylon 11,12** (Figure 19.4), as well as the urethane-based systems mentioned previously. These polyamide resin dentures are often referred to as flexible dentures and they are claimed to have certain advantages, notably the absence of metal clasps, better processing speed, reduced cost, and greater impact strength than conventional RPD materials. These flexible materials are clearly advantageous for patients suffering from acrylic allergy and those that object to a metallic taste, whereas claims of toughness, nonbreakability, flexibility, improved esthetics, and comfort are obviously appealing. Further, flexible denture base materials are increasingly used for night guards, temporomandibular joint (TMJ) splints, space maintainers, gum veneers, and stabilization of periodontally compromised teeth.

RPDs (and complete dentures) fabricated with these thermoplastic materials are injection molded and they may be used in conjunction with metal frameworks. The physical properties of several flexible partial denture materials are summarized in Table 19.3.

Despite the claimed advantages of flexible RPDs, many patients dislike the apparent lack of denture stability during mastication and deglutition. Further, because of the lower strength and stiffness of these resins compared with metals, greater bulk is required in certain areas, typically for the clasps to abutment teeth and in high stress areas. This increased bulk is disliked by many patients and is not favored by clinicians because of associated problems that can arise with regard to abutment teeth. Regardless of these considerations, there appears to be a growing market for these flexible denture materials, not only in developing countries but also within the United States and Europe.

Most prosthodontists and dentists do not recommend the routine, continuous clinical use of flexible dentures because of problems with the flexibility of the integral clasps. However, these materials are useful in certain clinical situations such as when there are sharp undercuts on abutment teeth that prevent the use of conventional metal clasps and, particularly, for patients with acrylic or metal allergies. To date, there is limited information in the dental literature on the properties, clinical characteristics, and long-term success of these materials.



Figure 20.1 Maxillary complete denture. (Lucitone 199, courtesy of Dentsply International.)

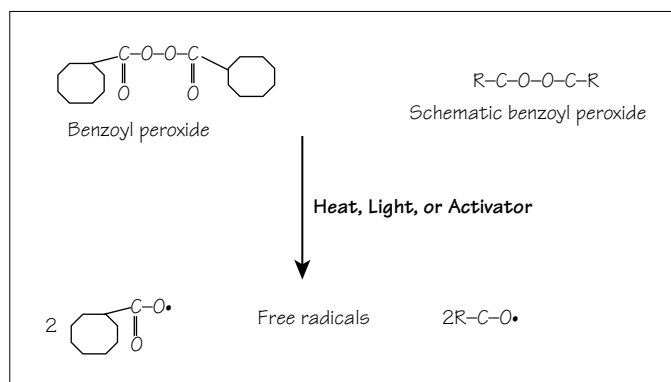


Figure 20.2 Reaction initiation—decomposition of benzoyl peroxide molecule into two free radicals.

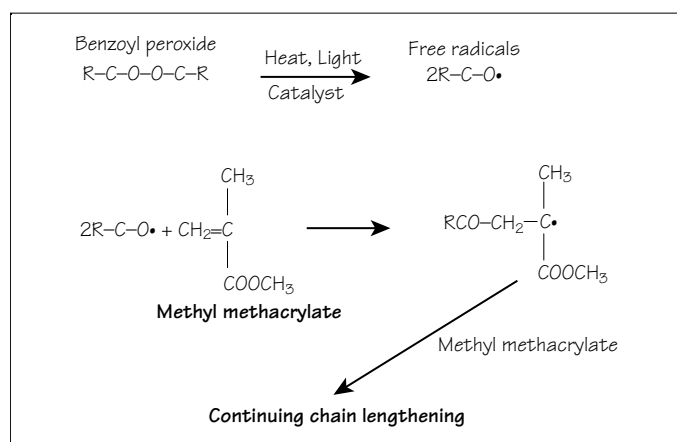


Figure 20.3 Reaction propagation—interaction between free radicals and monomer.

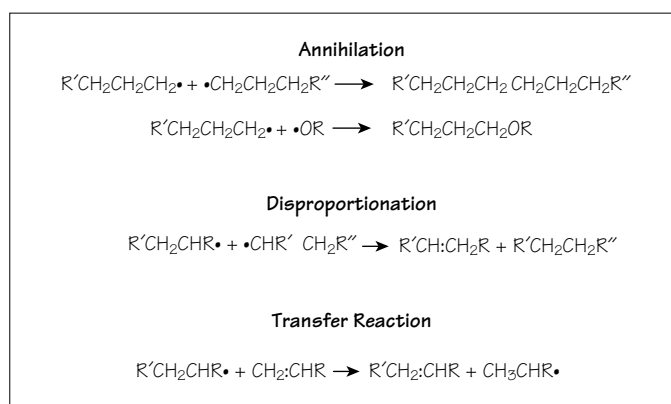


Figure 20.4 Reaction termination by annihilation, disproportionation, or transfer reaction.

Box 20.1 Required properties of denture base materials (in alphabetical order)

Absence of taste and odor
 Biocompatibility
 Bondable to resins, metal, and porcelain
 Chemical stability
 Color stability
 Ease of fabrication and repair
 Insolubility in oral fluids
 Long shelf life
 Low sorption of oral fluids
 Moderate cost
 Natural appearance
 Processing accuracy and dimensional stability
 Satisfactory thermal properties
 Strength and durability
 Wear and abrasion resistance

Table 20.1 Stages during monomer–polymer reaction

Stage	Mix characteristic
1	Sandy
2	Stringy
3	Doughy
4	Rubbery
5	Stiff

Table 20.2 Glass transition temperatures (T_g) of methacrylate resins

Methacrylate resin	Glass transition temperature, T_g (°C)
Methyl	125
Ethyl	65
<i>n</i> -Propyl	38
Isopropyl	95
<i>n</i> -Butyl	33
Isobutyl	70
<i>sec</i> -Butyl	62
<i>tert</i> -Amyl	76
Phenyl	120

Table 20.3 Residual monomer levels

Resin	Residual monomer (%)
Fully processed PMMA	≤ 2
Fluid or pour-type resins (Chapter 21)	2–5
Improperly processed PMMA	10–12

Table 20.4 Problems with denture bases and their causes

Porosity	Fracture
Loss of monomer	Biophysical and/or mechanical trauma
Inhomogeneous mix	Incomplete cure
Flask packed too early	Polymer characteristics
Insufficient dough in flask	Excessive heat or pressure during curing
Insufficient pressure on flask	
Discoloration	Distortion
Chemical effects	Processing stresses
Physical effects	Heat from grinding/polishing
	Relining
	Repairs
	Premature removal from flask

Complete dentures (CDs) are removable prostheses that replace the entire dentition. They consist of a continuous base covering the oral mucosa with attached prosthetic teeth (Figure 20.1). The requirements of denture base materials are summarized in Box 20.1. Although individual denture bases may be formed from metals, most CDs supplied to patients since the mid-1940s are fabricated from **poly(methyl methacrylate)** resin, also known as **PMMA** or **acrylic resin**. **Modified acrylics** and other resins are used for specific purposes and in response to new technology (see Chapter 21).

Denture base strength is determined by the polymer used, the processing regimen, denture shape and thickness, tooth positioning relative to the ridge crest, and the presence of any reinforcement.

20.1 Polymerization reactions

PMMA resin is formed by **addition polymerization** of **methyl methacrylate** by mixing polymer powder (partially polymerized methyl methacrylate) with monomer (methyl methacrylate), which then undergoes polymerization. When mixed, polymer powder and monomer liquid form a dough that is then packed into the mold or flask and, after setting of the teeth, is cured to form the denture. Note that methyl methacrylate (monomer) is toxic and must be handled with care; it can cause severe allergic responses in clinicians and patients.

1 Reaction initiation: Initiation requires the generation of **free radicals**, commonly released from **benzoyl peroxide** by chemical reaction, heat, or light (Figure 20.2). The chemical reactions involved in free radical formation include the following:

- benzoyl peroxide + aromatic **tertiary amine** + heat
- **camphorquinone** + tertiary amine + visible light

The tertiary amines include *N,N*-dimethyl-*p*-toluidine and *N,N*-dihydroxyethyl-*p*-toluidine. Sulfinic acids and their salts accelerate the reaction and are required for room-temperature curing (“**cold-curing**” or “**autocure**”) materials.

2 Reaction propagation: Free radicals react with centers of **unsaturation** (double bonds) leading to chain lengthening (Figure 20.3). Molecules such as **glycol dimethacrylate** in the monomer permit **cross-linking** to occur.

3 Reaction termination: Termination occurs by **annihilation**, **disproportionation**, or **transfer** (Figure 20.4). Annihilation is when two growing chains interact to form one large chain and propagation ter-

minates or when the growing chain interacts with the initiator. Disproportionation is when two growing chains interact, whereas a transfer reaction is when growth of the original chain stops and another shorter chain starts propagating.

4 Monomer must thoroughly wet the powder; after mixing, the mass is kept in a sealed jar to avoid monomer evaporation and to avoid toxicity and allergenic problems. Incomplete mixing results in streaked or blanched areas due to incomplete polymerization. The average working time of mixed powder and monomer is usually about 1 hour; on standing, mixed material goes through five reaction stages (Table 20.1). The **doughy** stage, lasting for about 1 hour, is optimal for packing the flask; packing at stages 1 or 2 results in excess free monomer and too fluid a mix; at stage 4 the mix is too viscous, resulting in incomplete flask closure, lost detail, and increased vertical dimension while teeth may drift or fracture.

20.2 Polymer properties

Properties are determined by the molecular weight (MW) of the **mer** (unpolymerized molecule) and by the polymer chain length, degree of chain branching, cross-linking and cross-link density within the molecule, residual monomer levels, and the presence of **plasticizers** and/or **fillers**. Plasticizers such as **dibutyl phthalate (DBP)** are added to produce a softer, more resilient polymer and to modify the **glass transition temperature (T_g)**.

1 Glass transition temperature: T_g is the temperature at which the resin transforms from a rigid material to one with rubbery characteristics and depends on such factors as the type of mer (Table 20.2). Generally, higher T_g values are found with more rigid chains, bulky substituents, and cross-linking.

2 Residual monomer: The level of residual monomer (Table 20.3) in the cured material adversely affects properties and can weaken the resin.

20.3 Problems with denture bases

Denture problems include porosity, distortion, crazing, fracture, and discoloration. The problems and their causes are summarized in Table 20.4.

20.3.1 Palatal vault

Dentures with shallow vaults have lower fracture strengths than medium and deep bases at all thicknesses. Tooth positioning relative to the ridge can affect denture strength because of the effect on denture flexure under loading.

20.4 Denture teeth

Denture teeth may be porcelain or modified acrylics, the latter being cross-linked for strength and crazing resistance. Plastic teeth are tough, are fracture resistant, and have excellent esthetics but are abraded by dental enamel. Porcelain teeth are hard and resist wear and abrasion but are brittle and may click in use. Although acrylic denture teeth will bond to the denture base, porcelain teeth require mechanical retention aids; many laboratories also groove acrylic teeth to improve retention.

20.5 Soft dentures

“Soft dentures” with claims of greater comfort because of built-in cushioning are often advertised by some dentists and by dental laboratories that provide complete dentures to patients independent of dental practitioners (“denturists”). Accordingly, there is a demand for soft dentures based on perceived greater comfort but these prostheses do not represent a new technology, rather, they are usually traditional hard acrylic denture bases with a soft (resilient) liner applied to the fitting (intaglio) surface of the denture base. Denture liners are discussed in Chapter 23.



Figure 21.1 PalaXpress Ultra overdenture. (Courtesy of Heraeus Kulzer US.)

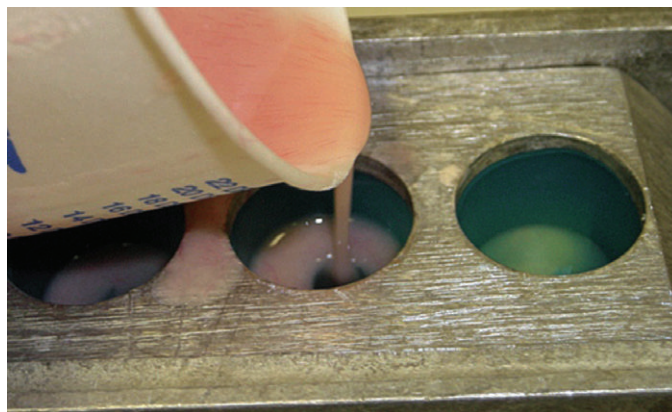


Figure 21.2 Pouring mixed pour resin and liquid into a mold. (Courtesy of Dentsply International.)



Figure 21.3 As-molded poured resin denture. (Courtesy of Dentsply International.)



Figure 21.4 Processed and polished poured resin Lucitone® Fas-Por™ denture. (Courtesy of Dentsply International.)



Figure 21.5 Triad® light-cured denture base with clear palate. (Courtesy of Dentsply International.)

Table 21.1 Properties of denture base resins

Resin	Knoop hardness (MPa)	Transverse strength (MPa)	Elastic modulus (GPa)	Impact strength (J/m ²)
Conventional	170	80–95	3.8	15–17
Autocure	160	84	1.6	15
Rapid cure	150–170	79–86	1.3–1.6	12–15
High impact	140	78–95	2.4–2.5	24–31
Light cure	175	110–120	4.2	25

Various modifications of poly(methyl methacrylate) resins are used for denture bases—some simplify processing whereas others provide certain characteristics to the denture base. Resin properties are summarized in Table 21.1.

One of the new generation of denture base polymers with hydrophilic properties is based on a biocompatible resin, poly(hydroxyethylmethacrylate), and has the trade name Hydrocryl®. This material has superior wetting characteristics compared with acrylic so that saliva and oral fluids will sheet on the intaglio (tissue contacting) surface of the denture, thereby improving retention. Further, the greater water sorption of poly(hydroxyethylmethacrylate) results in greater expansion, which compensates for polymerization shrinkage during processing and improves the accuracy and fit of the denture. This resin is processed by conventional laboratory techniques and has superior impact resistance. The material, however, is based upon a methacrylate resin and patients with methacrylate allergies may exhibit local reactions of mucous membranes.

21.1 High-impact resins

High impact acrylic resins, specially formulated acrylics, or **elastomers** such as butadiene-styrene rubber spheres are added to the powder (5–10%) to increase impact resistance. No special processing techniques are required, and many products are cold-curing and may also be poured and injection-molded. **Urethane dimethacrylate** materials modified with ethoxylated oligomers also are used and may contain rubber spheres. High impact resins are characterized by good color stability and increased toughness as well as improved fracture resistance compared with conventional acrylics (Figure 21.1)

21.2 Pour-type resins

The resin powder of fluid or pour-type resins comprises small particles that, when mixed with the liquid, typically monomer and 1,4-butanediol dimethacrylate, produces a fluid mix. The mixed material must be rapidly poured or injected into a hydrocolloid mold (Figure 21.2) and is polymerized under pressure (20–30 psi) in 106°F (41°C) water for about 15 min (Figure 21.3).

These resins may be weaker, have higher residual monomer content, and be more susceptible to distortion than conventional acrylics (see Table 20.4), although modern materials and techniques have eliminated many of these problems. Pour-type resins may also be injection-molded using customized processing units.

The overall processing time of a pour-type CD is markedly shorter than for traditional acrylic resins, and improvements in the appearance and properties of these resin dentures are continuously being made. Pour-type resins produce very acceptable complete dentures that can be characterized (Figure 21.4), and these materials also can be used to fabricate removable partial dentures and for the repair of fractured CDs.

21.3 Rapid-cure resins

The rapid-cure resins are **hybrid polymers** that incorporate both chemical and heat-activated initiators and that cure rapidly in boiling water. They generally are somewhat weaker than conventional acrylics (Table 21.1).

21.4 Light-cure resins

Light-cure systems are becoming increasingly popular because of speed and convenience; they usually are based on **urethane oligomers** and some comprise a **urethane dimethacrylate/acrylic copolymer** with microfine silica filler particles. They all contain a **photoinitiator** responding to blue light (400–500nm) and are processed in special units (see Figure 22.4). They have good mechanical properties but early systems tended to have greater water sorption and tended to stain more than conventional acrylic resins. Newer systems have eliminated these issues and all are monomer-free and, therefore, are kinder to the soft tissues.

The major advantage of light-cure systems is that they eliminate the need for the lost wax technique and do not require the investing or boiling out required with conventional acrylics. Many systems actually build up the denture using three layers—a base plate resin, a setup resin, and a contour resin—and, after try-in, the dentures are processed directly in custom curing units such that denture processing is simpler and faster than for conventional acrylic dentures. Further, the volumetric curing shrinkage is about 3% (compared with 7% for heat-cured acrylic resin) so that the fit of the final denture is improved and the processed material has a greater flexural strength and impact resistance than acrylic (Table 21.1). These urethane-based resins have good esthetics (Figure 21.5) and are inherently stain-resistant while also having a plaque resistance equal or superior to that of acrylic resin. These materials also are used for provisional restorations, RPDs, and night guards.

Light-cured systems do require custom light-curing units and because there is no inherent bond between the urethane resin and denture teeth, mechanical retention is required even for resin-based teeth.



Figure 22.1 Modern cold-cure denture repair material. (Courtesy of Lang Dental.)

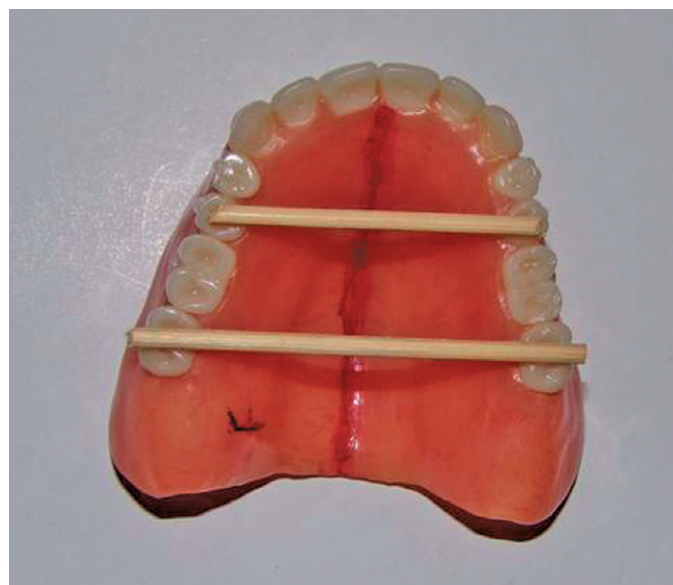


Figure 22.2 Arrangement of a fractured denture for repair.

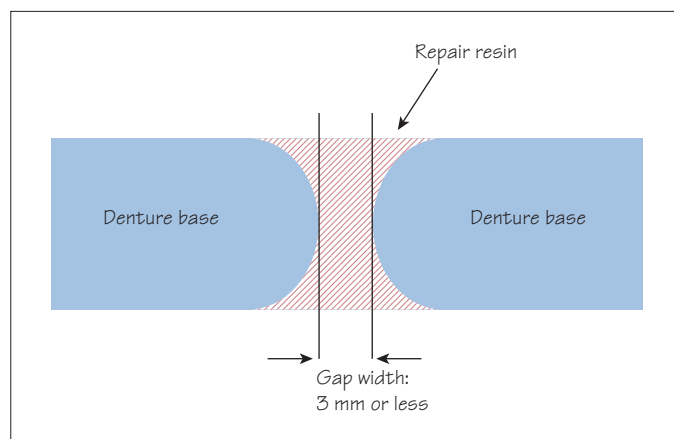


Figure 22.3 Fractured denture edges prepared with round profiles and a gap of 3 mm between broken edges for optimal repair.



Figure 22.4 Triad® custom curing unit for urethane dimethacrylate repair material. (Courtesy of Dentsply International.)

Dentures can fracture from a variety of causes, commonly through dropping onto a hard surface although other causes include failure of inherent defects from poor processing. Other factors predisposing complete dentures (CDs) to failure are a shallow palatal vault combined with severe flexure such as biting down onto a hard object or a deep incisal notch acting as a stress riser.

The commonest maxillary denture fractures occur at the midline from trauma (dropping) or arise from poor fit or occlusion problems. Although a repair can return the denture with a midline fracture to functionality, causative problems such as poor fit or unbalanced occlusion will remain and fracture can reoccur unless the clinical problems are addressed.

The advantage of acrylic resin as a denture base material is its ease of processing and tinting, ready polishability, and, in the present context, ease of repair. Formerly, denture repairs had to be made with heat-curing resin but modern cold-cure materials (Figure 22.1) are very satisfactory and their use reduces the risk of denture distortion during repair. The more recently introduced light-cured materials simplify the repair process. Because these materials are based on urethane oligomers, they are monomer-free and there is less risk of allergic reactions to the repair material. In addition, these materials are claimed to exhibit less fluid sorption and increased stain resistance than conventional cold-cure acrylics.

Regardless of the cause of the fracture or the mode of repair, the criteria for a satisfactory repair are clear (Box 22.1). Possibly of greatest importance are the requirements of rapid processing and maintenance of dimensional accuracy (i.e., “fit”) of the denture during and

Box 22.1 Criteria for a satisfactory denture repair

Repair process must be rapid.
Repaired structure must have adequate strength.
Denture dimensional accuracy must be maintained.
The repair must be esthetic.

after repair. A quick repair ensures that the patient is without the denture for the shortest period possible. Dimensional accuracy of the repaired prosthesis is necessary if the denture is to have the same functionality, fit, and occlusal balance as that prior to fracture.

A further factor related to denture fracture and repair is that the transverse strength of an acrylic denture base resin will decrease from about 950 kg/cm² (93 MPa) to ≤500 kg/cm² (49 MPa) after repair, regardless of repair methodology. There is a corresponding decrease in flexural resistance and impact strength. Thus, after fracture and repair, the denture will be some 50% weaker than prior to fracture although there are occasional claims that repairs can achieve up to 75% of the original transverse strength.

22.1 Denture repair

The key to a successful denture repair is accurate reassembly of the broken parts. The broken pieces are brought into apposition and maintained in position with sticky wax or a high-strength resin adhesive. Polymeric, metal, or wood rods then are attached to the occlusal surfaces of the teeth in opposing quadrants across the midline of the denture (Figure 22.2). Thereafter, a cast is poured with plaster or suitable die material (see Chapter 8 and Chapter 9).

After removal from the cast, the denture pieces are prepared for repair. A number of edge profiles have been suggested but studies indicate that a round profile (Figure 22.3) is the most convenient and yields excellent strength repairs. The traditional flat edge profile or butt joint provides a much weaker repair joint. A gap or separation of ≤3 mm between the broken pieces appears to be optimal.

Research suggests that polishing the broken edges with pumice followed by treating with monomer (methyl methacrylate) for 2 minutes results in repairs of greatest flexural strength and fracture toughness. After preparation of the edges, the denture parts are placed back on the cast and the repair is built up using alternate applications of monomer and powder, the so-called **salt-and-pepper** technique. The repair material is applied only to the polished (cameo) surface of the denture and should be slightly overbuilt to allow for finishing.

Repair is accelerated, and porosity in the repair resin is reduced, by placing the denture in a pressure-curing unit just covered by warm (100°F or 38°C) water. The applied pressure should be about 30 psi (0.2 MPa) and curing performed for at least 10 minutes. After removal of excess repair resin, the denture can be polished and returned to the patient.

Maintaining a separation of 3 mm or less ensures that if fracture occurs again, the failure line will run through the middle of the repair area rather than at the interface between repair material and the denture. This ensures greater strength for the repaired denture.

The use of light-cured materials, e.g. Triad, can simplify and speed up the repair process although the criteria of edge preparation and careful alignment of the broken denture pieces must be observed. The advantage of these light-cured materials is that they can be used to “tack” the broken pieces together with a hand-held curing light prior to filling the gap between them. The final repair is accomplished in a few minutes but requires the use of a custom curing unit for a satisfactory repair (Figure 22.4). The final repair and the strength of the repair material are claimed to be stronger than that achievable with acrylic resin.

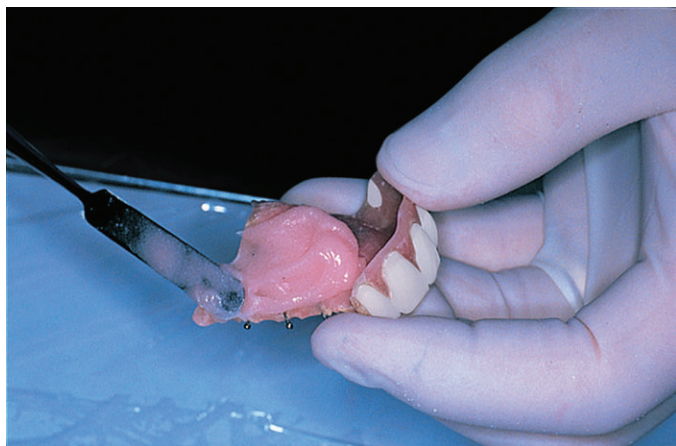


Figure 23.1 Placement of PermaSoft within a denture. (Courtesy of Dentsply International.)

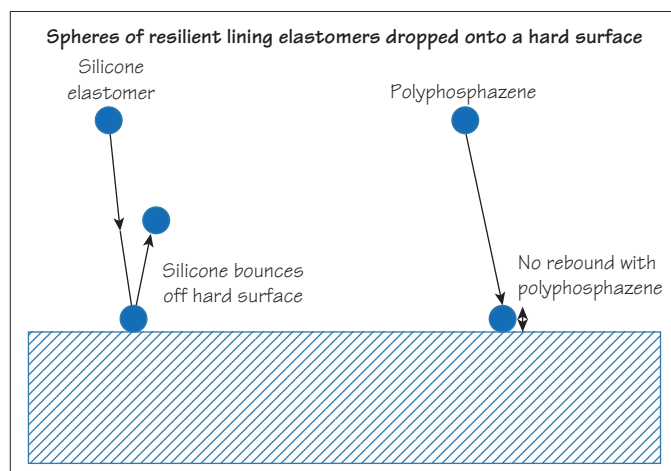


Figure 23.2 Schematic representation of shock-absorbing capability of silicone and polyphosphazene elastomeric resilient liners.

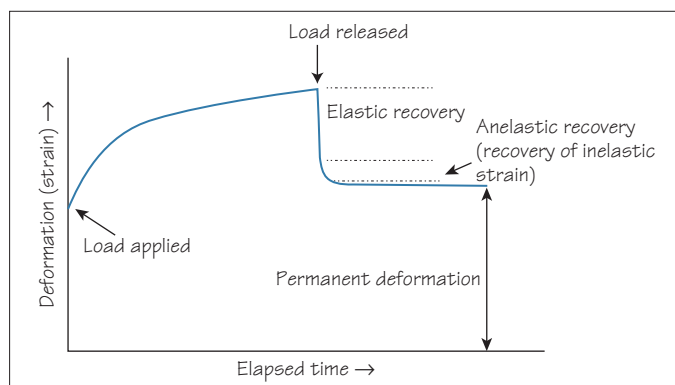


Figure 23.3 Elastic, anelastic, and permanent deformation of a viscoelastic gel under loading and after load release.

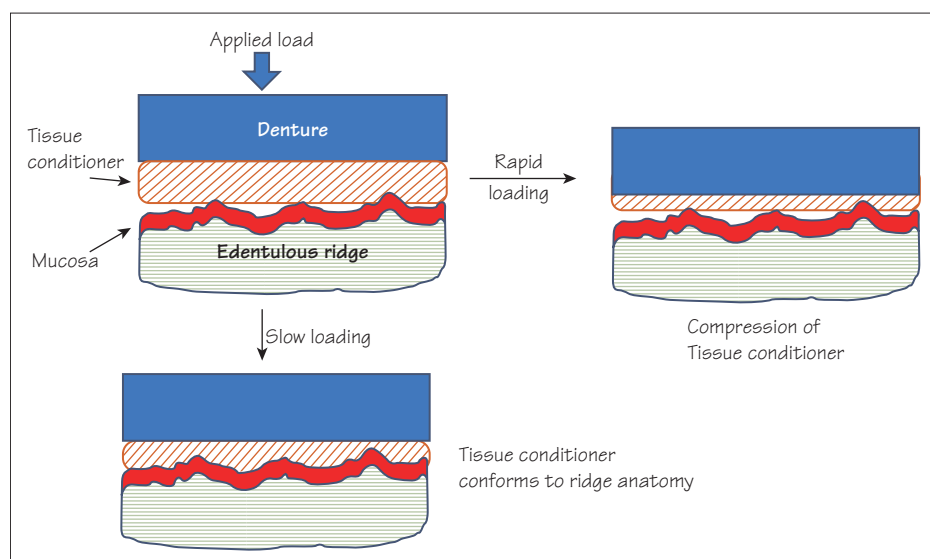


Figure 23.4 Schematic representation of the deformation of a tissue conditioner under rapid (dynamic) loading and slow (quasi-static) loading.

Hard denture relining with PMMA or light-cured materials improves fit or corrects occlusion problems whereas **soft** (or resilient) liners are used as cushioning materials placed on the intaglio surface, as noted in Chapter 20. Denture liners fall into two broad categories, **resilient liners** and **tissue conditioners**.

23.1 Resilient (soft) liners

Soft liners provide “shock-absorbing” (cushioning) by evenly distributing functional loads and avoiding local stress concentrations, thus helping manage traumatized oral mucosa, bony undercuts, bruxism, ridge atrophy, and congenital oral defects. The principal resilient liner materials are plasticized acrylics, vinyl polymers, and silicone and polyphosphazene elastomers. Figure 23.1 shows placement of a resilient liner within a denture.

23.1.1 Resilient liner materials—modified acrylic resins

Because of their free monomer content, conventional acrylics are contraindicated for direct placement on irritated mucosa, even when heavily plasticized. Acrylic-based resilient liners commonly combine poly(ethylmethacrylate) (PEMA) and methyl methacrylate monomer together with a plasticizer such as butyl phthalyl butyl glycolate (BPBG) or dibutyl phthalate (DBP).

PEMA has a T_g of 65°C, lowered by plasticizer addition, compared with 125°C T_g for PMMA, and their chemical similarity facilitates bonding to the denture base. The liner may be autopolymerized directly onto the denture base (chairside relining) or may be laboratory processed. DIY reliners are commonly based on ethyl methacrylate polymerized by benzoyl peroxide. They are plasticized by ethanol and 2-ethylhexyl diphenyl phosphate (EDP).

Plasticizer leaching and progressive absorption of saliva over time cause gradual liner hardening. Bacterial and fungal colonization also occurs, rendering the liner unacceptable.

Monomer-free chairside temporary resilient liners, based on PEMA plasticized with BPBG or DBP and ethanol, bond relatively poorly to the denture base. Resilient liners based on higher methacrylates, e.g. ethyl, *n*-propyl, and *n*-butyl, have lower T_g s and need less plasticizer, thereby reducing leaching.

Vinyl resin resilient liners, typically based on poly(vinyl chloride), poly(vinyl acetate), and vinyl acrylics, are also used. These materials likewise leach plasticizer over time and harden progressively.

23.1.2 Resilient liner materials—silicone elastomers

Silicone liners are soft at or below mouth temperature, with both autopolymerizing and heat-curing types available. Typically based on α - ω -dihydroxy end-blocked poly(dimethylsiloxane), they set by condensation polymerization. Chairside (cold-curing) silicone resilient liners are addition-curing vinyl polysiloxanes possessing dimensional stability and good elasticity retention.

Silicone liners bond poorly to PMMA, requiring bonding agents such as a silicone polymer (e.g., γ -methacryloxypropyltrimethoxysilane) solution, or cements based on rubber–poly(methyl methacrylate) mixtures. Silicone copolymers have also been developed that incorporate components that provide bonding between liner and denture base.

Formerly, cold-curing silicone liners had poor mechanical properties and exhibited significant volumetric change with sorption or loss of water that affected liner–base bonding. Newer polyvinyl siloxanes, however, have better long-term stability and superior elastic proper-

ties. The heat-cured materials are more stable in the mouth and may constitute “permanent” soft linings.

The newest advance in permanent resilient liners is Novus®, a polyphosphazene. This material has exceptional energy-absorbing properties (Figure 23.2), contains no plasticizers, and bonds well to PMMA. Readily processed within the dental laboratory, polyphosphazene does not harden, wets readily, and resists surface and subsurface fungal growth.

23.2 Tissue conditioners

Tissue conditioners are cold-cured viscoelastic gels considerably softer than resilient liners. The resin monomer may be methyl methacrylate or a mixture of methyl methacrylate and 2-hydroxyethyl methacrylate; 1,3-butanediol dimethacrylate facilitates cross-linking. Material resilience is due to heavy plasticization, typically with 40–55% DBP or BPBG.

Tissue conditioners, commonly chairside reliners, are used for irritated mucosa beneath dentures. When applied to a denture and the jaws brought into occlusion, the gel conforms to the anatomy of the residual ridge. Under rapid or dynamic loading (e.g., mastication), the material behaves elastically and resists flow, but under constant (static) or slow loading, the tissue conditioner flows, conforms to the underlying ridge, and redistributes stress (Figure 23.3 and Figure 23.4).

23.3 Problems with denture liners

23.3.1 Useful life

Tissue conditioners are temporary treatment modalities, requiring regular replacement; both residual monomer and plasticizers are leached over time, about 50% plasticizer loss occurs within 24 hours, the remainder leaching in 3–5 days. Plasticizer loss causes hardening and necessitates regular replacement, commonly in 3–5 days although they may last up to 3 months.

Resilient liners can survive for 6 months, and sometimes up to 1 year, before requiring replacement. More recently, liners containing a polymerizable plasticizer that becomes incorporated within the liner have been reported; this development should reduce plasticizer leaching and extend liner service life. Silicone and polyphosphazene liners have the greatest clinical durability.

23.3.2 Liner thickness

To be effective, resilient liners should be 2–3 mm thick, requiring a corresponding reduction in denture base thickness to maintain denture fit and occlusion. Such thinning will weaken the base and, additionally, materials used with liners (adhesives and monomers) can attack the denture base, exacerbating strength loss. This promotes susceptibility to fracture in use or if the denture is dropped.

23.3.3 Liner separation

Liner–base separation and liner breakdown can occur over time, the latter causing fissuring, degradation, and roughening, allowing ingress of food debris and bacteria into the liner and crevices between base and liner. Micro-organism proliferation over the fitting surface will cause an unpleasant taste and denture odor.

23.3.4 Liner sanitization

Liners and tissue conditioners require regular cleansing to eliminate bacterial accumulation, fungal growth, and general staining but their softness contraindicates mechanical cleaning. Oxygenating cleansers are useful but their success can be variable, particularly with silicone materials, which may be susceptible to damage by these cleansers.

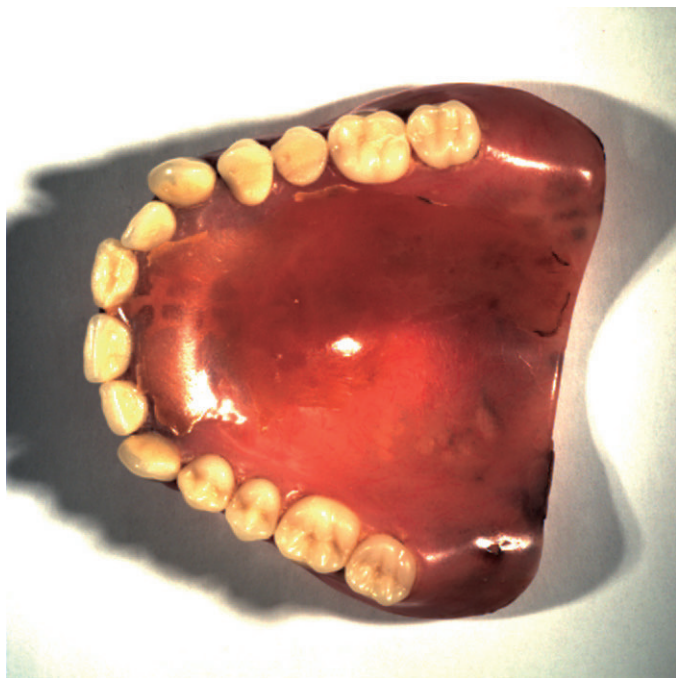


Figure 24.1 Cameo surface of a dirty denture. (Courtesy of GSK.)

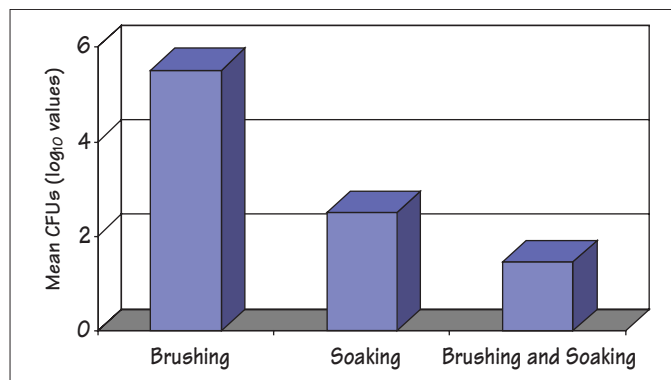


Figure 24.2 Numbers of colony-forming units (CFUs) recovered from dentures after cleaning. (Based on Dills SS, Olshan AM, Goldner S, Brogdon C. Comparison of the antimicrobial capability of an abrasive paste and chemical-soak denture cleaners. *J Prosthet Dent* (1988) 60(4):467–470.)

Box 24.1 Bacteria and fungi identified from dentures

Gram-positive organisms

Staphylococcus species
Streptococcus species

Gram-positive rods

Arcanobacterium haemolyticum
Actinomyces species

Gram-negative rods

Pseudomonas aeruginosa
Pseudomonas fluorescens
Burkholderia cepacia
Stenotrophomonas maltophilia
Enterobacter cloacae
Klebsiella pneumoniae

Gram-negative cocci

Neisseria perflava

Fungi

Candida glabrata
Candida albicans
Candida paratropicalis

(Based on Glass RT, Bullard JW, Hadley CS, Mix EW, Conrad RS. Partial spectrum of microorganisms found in dentures and possible disease implications. *J Am Osteopath Assoc* (2001) 101:92–94.)

Denture cleanliness and bacterial colonization are strongly correlated. The prevalence of denture stomatitis, reportedly as high as 65%, emphasizes the importance of early prevention of binding and accumulation of denture plaque. Denture materials do not resist adherence and possible damage by oral bacteria, and soft liners, tissue conditioners, and denture adhesives are particularly susceptible to microbial growth. A typical dirty denture is shown in Figure 24.1; however, there is limited information in the literature on effective denture cleansing.

Effective cleansing requires both the rapid and efficient removal of debris, stains, denture plaque, and any bacterial colonization. The cleansing agent, however, must not cause damage to the denture, liner materials, or other appliances nor can it leave toxic, malodorous, or unpleasant-tasting deposits.

24.1 Denture plaque

Substantial microorganism contamination of dentures (ca. 10^4 – 10^6 CFU/cm²) occurs within 24 hours, with posterior regions being more heavily contaminated than anterior. Further, denture interiors are generally more contaminated than surfaces due to their inherent porosity, permitting microbial permeation and contamination throughout the entire denture. Microorganism colonization causes the formation of deposits (denture plaque); the role of plaque in persistent or recurrent stomatitis has been recognized for over 40 years.

An extensive spectrum of opportunistic and pathogenic microorganisms colonizing dentures has been identified (Box 24.1). The hundreds of bacterial species in the oral microflora can cause substantial oral infections and may induce systemic diseases; certain of the isolated bacteria, e.g. *Klebsiella* and *Enterobacteriaceae*, may play a role in denture malodor.

24.2 Cleansing methods

Two common approaches to denture cleansing are brushing with a toothbrush and dentifrice and immersion in a chemical cleaning agent. Commercial (household) cleaners such as Clorox and Calgon are effective fungicides *in vitro* but not under clinical conditions. Likewise, chlorine dioxide reduces, but does not eliminate, viable microorganisms on dental prostheses and, although hypochlorites are effective stain removers, they can cause surface deterioration. Therefore, household cleaners are not only ineffective but may cause damage to prosthetic materials.

24.2.1 Brushing of dentures

Denture cleaning by brushing with dentifrice is popular but has limited effectiveness in removing microorganisms whereas there are associated problems. Irrespective of bristle stiffness, there is negligible abrasivity from brushing with water alone but abrasion can occur with dentifrices. In particular, dentifrices and even low-abrasivity pastes can cause roughening and grooving of dentures and liners, with the

cleanser, brush characteristics, and the denture material affecting surface abrasion. Further, toothbrush bristles flex during use and can bring more abrasive particles into contact with the surface, enhancing the abrasive and cleansing action but also increasing the risk of surface damage. Surface grooving from abrasion increases denture and liner susceptibility to stain buildup and plaque accumulation.

Greater denture cleaning efficiency is possible when both an immersion cleanser and brushing are used. This is shown by the reduced numbers of colony forming units (CFUs) recovered from dentures after cleaning (Figure 24.2).

24.2.2 Effervescent cleansers

Immersion cleansing is effective and convenient for plaque control, particularly for soft liners. This is an important consideration because tissue conditioners and soft liners support mycotic colonization and growth, leading to fungal penetration into these materials.

Immersion cleansers are based upon monopersulfate and perborate oxidants. These agents decompose to release oxygen, which provides surface cleansing and antimicrobial activity against gram-negative anaerobic rods, gram-positive facultative cocci, and gram-negative anaerobic cocci as well as *Streptococcus mutans*. There have been a few reports that regular cleanser use causes whitening of dentures, chairside relines, and appliances fabricated from autopolymerizing acrylic resins, the latter being less stable than heat-processed materials. It also is possible that the peroxide content and alkaline pH of effervescent cleansers may contribute to increased surface roughening of liners, oxygenation in strongly alkaline solution being the primary damaging factor. It has been suggested that dentures fabricated from light-activated resins may be optimal for patients prone to denture stomatitis as they show the least overall degradation from candidal treatment modalities.

Novel cleansers containing enzymes such as β -1,3-glucanase, mutanase, and protease show significant *Candida*-lytic activity and appear effective in removing denture plaque. This may be due to their ability to destroy intercellular adhesion, resulting in fungicidal activity and the ability to remove yeasts.

Although immersion cleansers are superior in cleaning action to denture pastes, they may not be completely effective in controlling heavy plaque when used alone in that there is only 30% removal of the heavy plaque that can accumulate within 3 days on a denture without adequate hygiene. Conjoint use of ultrasonic agitation and immersion cleansing, however, reportedly removes 75–78% of the plaque although little is known of the effects of ultrasonic agitation on denture base materials, particularly those with resilient liners or tissue conditioners

Although microwave-assisted denture cleansing may be effective in plaque removal, the heating effects associated with microwave radiation may be a problem, particularly with lower T_g materials.

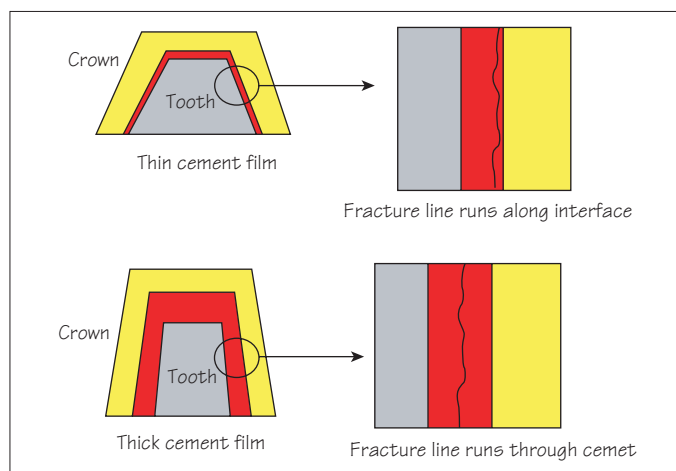


Figure 25.1 Failure patterns for thin and thick cement films beneath crowns. (From *Fundamentals of Fixed Prosthodontics*; M. Palermo and J.A. von Fraunhofer, Wiley-Blackwell, 2013.)

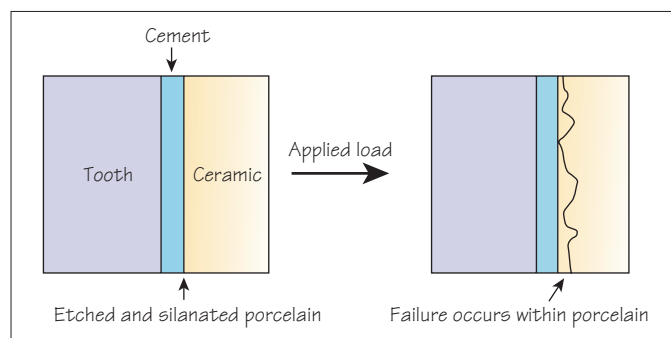


Figure 25.2 Bond strength of adhesive to etched and silanated porcelain exceeds the cohesive strength of porcelain. (From *Fundamentals of Fixed Prosthodontics*; M. Palermo and J.A. von Fraunhofer, Wiley-Blackwell, 2013.)

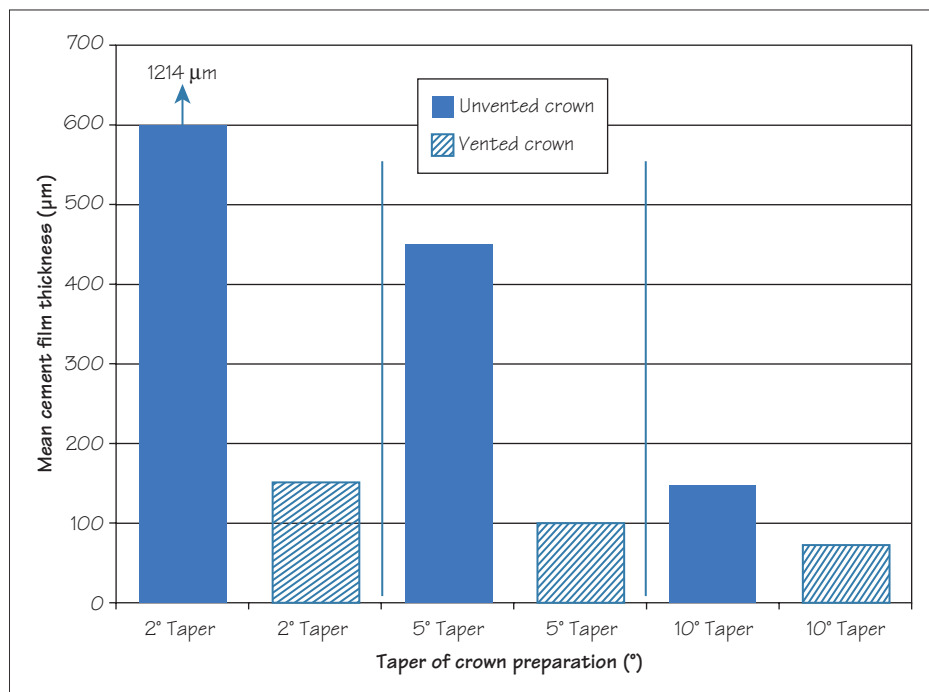


Figure 25.3 Cement film thicknesses in μm beneath vented and unvented full-coverage crowns for different taper angles.

Box 25.1 Cement selection criteria

Biocompatibility
Low thermal conductivity
Low thermal expansion
Low solubility
Low viscosity
Low film thickness ($<20\mu\text{m}$)

High cohesive strength
Good adhesion
Opacity
Long working time, snap set
Easy manipulation
Long shelf life

Technically, cements are substances that “set,” but **dental cements** are luting agents or “lutes” used to bond prostheses to prepared teeth or implants. Thus dental cements are bonding agents (adhesives) that hold prostheses in place through strong interfacial forces.

Three broad categories of dental cements exist: those providing a protective action (liners and bases, Chapter 26); those that are temporary (provisional) luting agents (Chapter 27); and those for permanent luting of restorations (Chapters 28–30). Traditionally, dental cements

were inorganic zinc oxide–based materials but now resin-based materials predominate.

Ideally, cements should wet both the hard tissue substrate and the prosthesis, and provide either a strong chemical interaction or mechanical interlocking with both surfaces. Additionally, the cement should possess long-term stability in use, i.e. exhibit good cohesion (high internal strength); see Box 25.1.

The adhesive layer between restoration and tooth should be as thin as possible so that the cement–tooth and cement–restoration interfacial adhesions exceed the cement cohesive strength, maximizing the shear force required to overcome interfacial retentive forces. Where there are undercuts or poor fit between restoration and tooth, the cement layer thickness increases to fill the gap during restoration placement and the luted system can fail under lower applied forces. During failure, cracks generated by the applied force no longer have to shear through adhesive forces and mechanical interlocks between cement and substrate, but propagate through the much weaker cement layer, overcoming its cohesion (Figure 25.1). The net result is that thicker cement films equate to reduced retention.

25.1 Dental cements—general considerations

Presently, the perfect cement does not exist. For example, a highly adhesive luting agent with good cohesive strength and a snap set appears ideal but, during crown cementation, excess luting agent may be squeezed out and then set rapidly. Removal of excess material might be difficult and, if left *in situ*, would constitute artificial calculus, causing periodontal problems.

Based on their chemistry, cements are broadly classified as inorganic (e.g., zinc phosphate cement), organic salt (e.g., zinc polycarboxylate and glass ionomer cement), and resins or resin-reinforced adhesives. Inorganic and organic salt cements rely on mechanical interlocking into surface discontinuities and roughness although, with organic salt cements, there is some chemical bonding between the organic acid and calcium in hard tissue. In contrast, resin cements rely more upon chemical interactions for their retentive action. Thus, until the advent of resin-based cements, retention by dental cements was primarily mechanical in nature.

25.2 Cement properties

Biocompatibility is essential and no adhesive should cause tissue necrosis or elicit allergic reactions. In cement selection, certain properties are emphasized over others, typically consistency and **flow (viscosity)**. For adequate crown retention, the cement must flow readily over both tooth and restoration, forming a thin, strong, coherent, and void-free layer between them. Cement flow is determined by the inherent characteristics of the adhesive and the **P/L**, or mixing ratio, both determining film thickness.

Because all metallic restorations have greater thermal conductivities than hard tissue, dental cements must have low conductivity to provide a thermal barrier beneath restorations. Likewise, cement thermal expansion and contraction should not exceed that of tooth structure or restorative materials.

Mechanical properties likewise are important. The dental cement must possess sufficient strength and stiffness to provide support for the restoration and withstand masticatory forces. The importance of cohesive strength was discussed previously.

Depending upon restoration accuracy and fit, gaps of variable width always exist at the exterior and interior line angles of restorations.

Accordingly, cements should have low solubility to avoid dissolution in the event of oral fluid ingress into these gaps.

Long working times followed by a rapid or snap set are desirable characteristics but, as noted, they can cause problems. Also, use of rapid-setting adhesives requires accurate restoration and veneer placement before setting occurs, to avoid problems in removing misplaced restorations.

25.3 Luting agent selection

The restoration partly determines luting agent selection. Visible light-cured (**VLC**) adhesives are obviously useful for thin, translucent veneers but are less satisfactory for thicker or more opaque ceramic materials, or for adhesive layer thicknesses >2 mm. In such cases, dual-cure systems assure better long-term bonding success.

The internal (tissue-contacting) surfaces of cast metal restorations are rough and do not need additional bonding aids, but smooth restoration surfaces (e.g., ceramics) require etching. In fact, resin-based adhesive bond strengths to properly **etched and silanated porcelain** exceed the ceramic cohesive strength so that fractures occur within the porcelain layer rather than at the porcelain–cement or cement–tooth interfaces (Figure 25.2).

The relatively new technique of **tribochemical bonding** is used for pretreating materials with high chemical resistance, such as zirconia. In this, a layer of 110- μm Al_2O_3 particles modified with silica is deposited on the ceramic surface by sand-blasting and is then silanated. Tribochemical treatment (i.e., alumina–silica coating and subsequent silane treatment) markedly improves bonding of self-adhesive resin cements to zirconia.

25.4 Cement film thickness

An important consideration in the clinical success of a luted restoration is the thickness of the cement film beneath that restoration. This cement film thickness is significant because it affects both the possible failure mechanism, as noted above (Figure 25.1) and the accuracy of the restoration fit. A thick occlusal cement film, for example, will prevent the restoration from being properly seated, resulting in greater cement thicknesses at the margins and potential clinical problems. In particular, a thick cement layer at the external line angles can be washed out, resulting in leakage and secondary decay as well as periodontal problems arising from excess subgingival cement for restorations with buried margins.

Although ADA/ISO specifications require that cements have film thicknesses of 25 μm or less, the actual thickness of the cement layer beneath a restoration can be as much as an order of magnitude greater than this value. In fact, an *in vivo* study showed that the average occlusal cement film thickness for clinically acceptable restorations was in the range of 100–150 μm , whereas that at the margins was 40–80 μm .

Several factors affect the cement film thickness beneath restorations, notably accuracy of restoration fit, flow properties of the cement, the preparation taper angle, and, in the case of full coverage crowns, whether the restoration is vented. One study, using simulated full coverage crowns fabricated under close engineering tolerances (± 2 μm), clearly demonstrated the effect of preparation taper and crown venting on cement film thickness (Figure 25.3). The data indicated that the greatest determinant of cement film thickness was the taper angle, although crown venting could dramatically reduce the occlusal layer of cement with steeper (i.e., near-vertical) tooth preparations. Comparable effects were found for the cement film thickness along the axial walls and gingival margins.

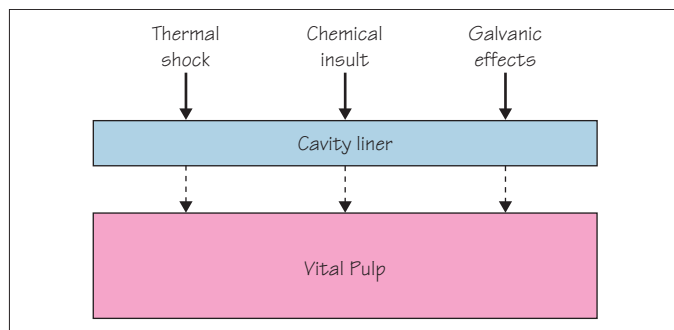


Figure 26.1 Cavity liner reducing the adverse effects of heat, chemicals, and galvanic effects on the dental pulp.

Box 26.1 Requirements of bases and liners

Biological compatibility
Low thermal conductivity
Low solubility
High strength
Chemical protection

Box 26.2 Factors affecting integrity and efficacy of liners and base

Degree of set at restoration placement
Strength and thickness of liner
Cavity design
Force exerted during restoration placement
Support by surrounding hard tissue
Operative technique

Box 26.3 Requirements of high-strength bases

Provide thermal protection
Support the restoration
Higher strength than liners
Higher viscosity than liners
Bond to tooth substance

Table 26.1 Average mechanical properties of low- and high-strength bases

	Compressive strength (MPa)	Tensile strength (MPa)	Elastic modulus (GPa)
Low-strength bases			
Ca(OH) ₂ —self-cure	15	1.0	0.4
Ca(OH) ₂ —VLC	96	38	—
Glass ionomer—self-cure	50	6	2.0
Glass ionomer—VLC	70	12.5	3.5
ZOE	5.5	0.4	0.3
Resin-based	65	—	2.3
High-strength bases			
Glass ionomer—self-cure	150	—	4.2
Zinc polyacrylate	80	16	5.0
Resin-based	190	—	7.5

Cavity varnishes can provide barriers at the tooth–restoration interface against migration of irritants from restorations and/or oral fluids into the dentin. Cavity liners function both as barriers and in exerting therapeutic effects.

26.1 Cavity varnishes

A **cavity varnish** provides a **barrier** against fluid penetration into the underlying dentin. When applied to dentin, a varnish may reduce postoperative sensitivity and limit migration of corrosion products into dentin, thereby reducing tooth discoloration.

26.1.1 Composition

Varnishes usually are solutions of natural gums, synthetic resins, or rosin (typically Copal and cellulose nitrate) with medications such as chlorobutanol, thymol, or eugenol added for therapeutic or obtundent effects. Solvents may be chloroform, alcohol, acetone, benzene, toluene, ethyl acetate, and/or amyl nitrate; solvent evaporation deposits a thin film on the substrate surface.

26.1.2 Varnish film characteristics

Films have minimal mechanical strength and provide no thermal insulation. Film thickness is 2–40 μm and the contact angle on dentin is 53°–106°, ensuring good wetting of clean and “dry” dentin surfaces.

26.1.3 Applications

Varnish films primarily exert a barrier effect but, because of their solvent content, they should not be used beneath resin-based restorations. They also may inhibit wetting of enamel and dentin by adhesives.

26.2 Cavity liners

Cavity liners provide both a barrier action (Figure 26.1) and therapeutic effects. They do not set per se but function more as film formers akin to varnishes.

26.2.1 Composition

Liners commonly are suspensions of **calcium hydroxide** [$\text{Ca}(\text{OH})_2$] in solvents such as methyl ethyl ketone, ethyl alcohol, or an aqueous solution of methyl cellulose, the latter functioning as a thickening agent. Liners may contain fillers (**PMMA** beads or barium sulfate) and a fluoride source such as calcium monofluorophosphate.

26.2.2 Characteristics

Cavity liners are mechanically weak and provide minimal thermal protection. The soluble film should not extend over margins. They have better cohesion and interact less with resins than varnishes; fluoride release reduces 2° decay and dentinal hypersensitivity. The requirements of liners and bases are summarized in Box 26.1; factors affecting liner integrity and efficacy are given in Box 26.2.

26.3 Low-strength bases

Low-strength bases are known as liners, intermediary bases, or pulp-capping agents. They set to a solid mass and commonly are two-paste $\text{Ca}(\text{OH})_2$ or ZOE cement systems.

26.3.1 $\text{Ca}(\text{OH})_2$ bases

The base paste contains calcium tungstate, tribasic Ca phosphate, and ZnO in glycol salicylate while the catalyst paste contains $\text{Ca}(\text{OH})_2$,

ZnO, and zinc stearate in ethylene toluene sulfonamide with fillers (calcium tungstate or BaSO_4) providing radiopacity. Setting occurs by reaction between $\text{Ca}(\text{OH})_2$ and salicylate to form amorphous calcium disalicylate.

26.3.2 ZOE bases

ZOE bases are basically ZOE cements (Chapter 27) and set through formation of zinc eugenolate, accelerated by moisture and temperature elevation.

ZOE bases are used in deep cavities to retard acid penetration and may be used in conjunction with high-strength bases.

26.3.3 Visible light–cured bases

VLC bases often are based on **urethane dimethacrylate** resins and contain barium glass and/or barium sulfate. Some products also contain dispersed sodium **fluoride** for caries protection.

26.3.4 Properties of low-strength bases

1 $\text{Ca}(\text{OH})_2$ bases: These bases are used to line deep cavities or for direct pulp capping. Setting time is 2½–5½ minutes with progressive strength increase over 24 h. Water solubility is low to moderate. They have a low elastic modulus and require sound dentin, but film thicknesses of ≤ 0.5 mm will withstand amalgam condensation and provide some degree of thermal protection. Surface pH is high, and unreacted $\text{Ca}(\text{OH})_2$ stimulates 2° dentin and provides antibacterial activity.

2 ZOE bases: These bases are used in thin layers and provide little thermal insulation. They have a lower strength and elastic modulus than $\text{Ca}(\text{OH})_2$. Eugenol content can inhibit polymerization of resin-based materials.

3 VLC liners: Light-cured liners are being used increasingly and generally exhibit good bond strength to dentin. They have high surface pH, low acid (0.5%) and water (1.0%) solubility, and high strengths but little antibacterial activity.

26.4 High-strength bases

High-strength bases have different requirements than liners and low-strength bases (Box 26.3); strength properties are summarized in Table 26.1.

1 Zinc phosphate and zinc polyacrylate cements: These were formerly used as bases but less so now.

2 Glass ionomer cements: GICs (both self-cured and light-cured) are popular bases. Light-cured materials contain **dimethacrylate oligomers**, which provide 50% greater strength to the set material.

3 Resin-based bases: These are commonly dual-cured, paste–paste composites, typically a **bis-GMA** resin with a hydrophilic diluent and fluoride-releasing glass filler; radiopaque barium glass and bonding agents are often present.

26.4.1 Characteristics of high-strength bases

High-strength bases are more viscous and flow less than low-strength bases; at thicknesses > 0.5 mm, they provide thermal protection for the pulp and mechanical support for restorations. Resin-based materials bond to dentin whereas GICs bond to enamel and dentin.

Recent studies indicate that there is less need for sealing of cavity preparations or the resin–tooth interface with adhesive dentistry,



Figure 27.1 Provisional restoration in single quadrant impression. (Courtesy of Dentsply International.)

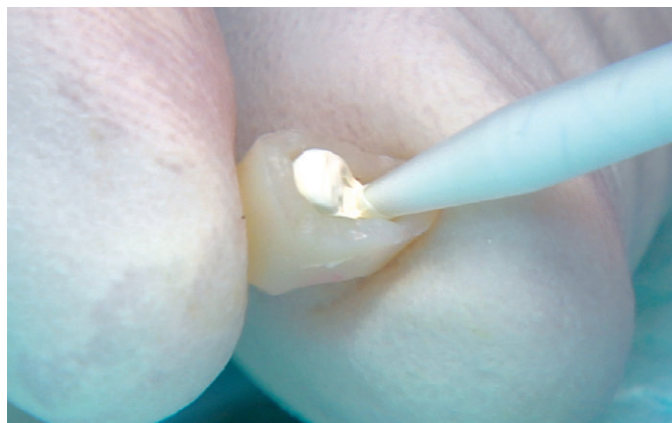


Figure 27.2 Loading Integrity® provisional restoration with provisional dental cement, prior to luting onto prepared tooth. (Courtesy of Dentsply International.)

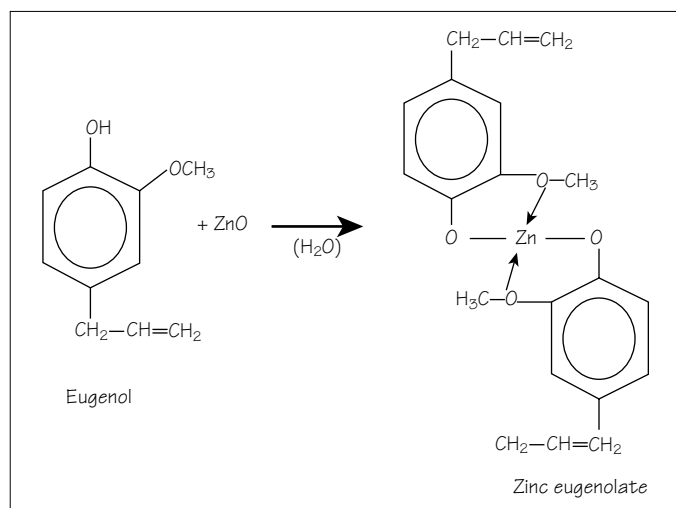


Figure 27.3 Setting reaction of zinc oxide–eugenol.

Box 27.1 Desirable properties of provisional cements

Biocompatibility with hard and soft tissues
 Easy dispensing, mixing, and application
 Adequate working and setting times
 Good retention
 Easy removal of excess cement from the external and internal restoration surfaces
 Easy removal of the restoration without damage to hard and soft tissues
 Minimal or no reaction with the restorative material
 No interference with adhesion of a final cement
 Good shelf life

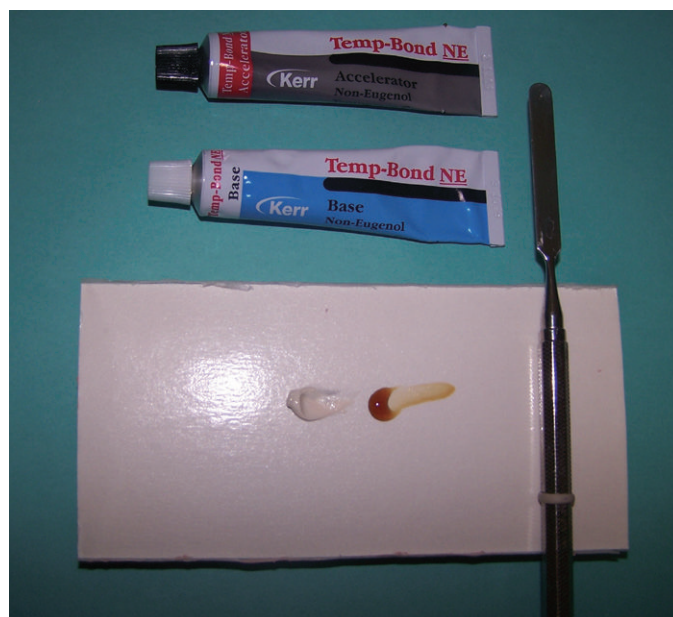


Figure 27.4 Dispensed paste-paste provisional cement. (Courtesy of Dr. Karen Hallisey.)

Table 27.1 Properties of provisional dental cements

	Working time (min)	Setting time (min)	Flow (mm)	Film Thickness (μm)	Compressive strength (MPa)	Transverse strength (MPa)
ZOE I	>1.5	<7	33	13	10	1
ZOE II	~1	<5.5	34	9	12	2
ZnO-NE I	1–1.5	2–3	27	11	22	9
ZnO-NE II	2	2.5	23	29	7	4
ZnO-NE III	>1.5	<7	32	8	6	3
ZnO-NE IV	~1	4–5	25	24	5	4
ZnO-NE V	>1.5	<5.5	36	17	41	2

Provisional or temporary cements are used for luting of provisional indirect restorations, including crowns, fixed partial dentures (FPDs), inlays, and onlays, as well as for temporary cementation of definitive (final) restorations. They are also recommended for the cementation of crowns and FPDs on implants. Desirable properties of provisional cements are summarized in Box 27.1. Provisional cements are also used for temporary filling of the access opening between root canal treatments, as interim restorations on implants, and during nonvital tooth bleaching. Typical clinical use of a provisional cement is shown in Figure 27.1 and Figure 27.2.

27.1 Zinc oxide–eugenol cements

Zinc oxide–eugenol (ZOE) cements were traditionally the luting agents for provisional restorations. They comprise a powder (ZnO) and liquid (eugenol) system, the setting reaction of which is initially hydrolysis of ZnO powder to zinc hydroxide, the Zn(OH)_2 then reacting with the methoxy group in eugenol to form an amorphous zinc eugenolate gel, usually at the surface of the ZnO particles (Figure 27.1). Over time, the gel crystallizes and increases the strength of the set mass; crystallite formation is accelerated by zinc acetate added to the ZnO . However, upon exposure to water (and saliva), zinc eugenolate breaks down, reverting back to Zn(OH)_2 with liberation of eugenol. The eugenol in ZOE was long thought to be an obtundent but, although it may have an anesthetic effect, it is in fact severely inflammatory.

The P/L ratio determines the consistency and mechanical properties of ZOE cements, which set in 4–10 min with a film thickness of $\leq 25 \mu\text{m}$ but the set material has low compressive strength (27.6 MPa). Modification with **orthoethoxybenzoic acid (EBA)** using a 1.66:1 EBA-to-eugenol ratio markedly increases cement strength.

ZOE cements are available as handmix (powder–liquid as well as paste–paste) and automix systems; the latter obviously are more convenient and provide greater accuracy in proportioning. As noted, ZOE is hydrolytically unstable—which, with its low strength, predicates its use as a provisional luting agent. However, eugenol release by hydrolysis of ZOE can leave residues that interfere with the setting reaction (polymerization) of subsequently applied resin-based permanent luting agents.

27.2 Noneugenol systems

Eugenol-free (or noneugenol, NE) provisional cements are now available and are supplied as paste–paste systems (Figure 27.4) or in automix syringes. These NE provisional cements are stronger, more retentive, and have longer working and setting times than ZOE but may have slightly inferior flow properties and somewhat greater film thicknesses. These cements are less susceptible to hydrolytic breakdown and, consequently, are considered to be more suitable for long-term provisional cements than ZOE.

Many NE materials were based on ZnO –poly–organic acid systems, typically polycarboxylic and polyacrylic acids. Setting involves a chelation reaction similar to that with eugenol. Many products also

contain sodium fluoride, tannin fluoride, and potassium nitrate for tooth desensitization and often antibacterials such as chlorhexidine and triclosan for preparation disinfection. Additives such as calcium phosphate and resins may be present to increase strength while some ZnO –carboxylate hybrid cements may contain some polycarboxylate resin to facilitate easy removal.

Although most of these cements are autocuring, dual-cure systems have been introduced and certain products have a two-stage curing reaction, namely, an initial gel-set that will stabilize the provisional restoration during removal of excess cement prior to a final, and more rigid, set. At least one new material is a single-component, moisture-activated, NE white cement designed for provisional cementation of crowns and bridges. Further, many products are supplied in regular set and more-rigid formulations for specific applications. Many NE cements are produced in tooth shades whereas others are translucent or have neutral shades so that there is no “show-through” in use. The absence of eugenol reduces interactions with composite restorations or resin-based permanent cements.

The properties of several ZOE and ZnO –NE provisional cements are summarized in Table 27.1. It can be seen that the properties of provisional cements with and without eugenol are similar, notably in their working and setting times as well as flow properties and film thicknesses. Whereas the transverse (flexural) strengths of ZnO –NE cements are generally greater than those of ZOE cements, only two ZnO –NE cements have greater compressive strengths than those of ZOE. In addition to the absence of eugenol, limiting adverse effects on subsequent permanent luting agents, ZnO –NE cements are claimed to have a smoother surface finish and to be retained within the restoration rather than the tooth when the provisional is removed. These cements are also claimed to ensure easier clean-up since they can be removed without crumbling.

Provisional cements based on glass ionomer technology (see Chapter 28) are also now available. Addition-cured, silicone-based zinc-oxide temporary cements with a silane agent for improved adhesion and marginal integrity have been introduced. These systems produce firm but elastic provisional cements that are easily peeled off the preparation and the internal surfaces of restorations.

27.3 Cement removal

It is important that all residual cement be removed from both the tooth preparation and the final restoration. Removal techniques include scraping the preparation with a scaler or curette, cleaning with a prophylaxis cup and water–pumice paste, and using an intraoral sandblaster. Of the three, intraoral sandblasting is the most effective whereas manual scraping is the least. Note that all surfaces of the tooth preparation must be cleaned, especially the margins. Likewise, all residual cement must be removed from a final (definitive) restoration that has been provisionally cemented, typically by sandblasting, ultrasonic cleaning, or rotary cleaning. Wiping all surfaces with ethanol will clean them but less satisfactorily. As noted above, ZnO –NE cements are claimed to offer easier removal from the tooth.

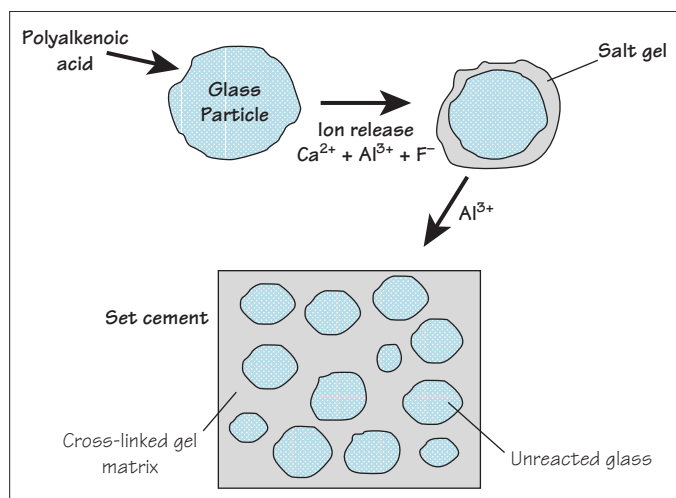


Figure 28.1 Setting reaction of glass ionomer cement (GIC).

Table 28.3 Zinc polycarboxylate and GIC characteristics compared with those of zinc phosphate (ZNP) cement

Characteristic	Zinc polycarboxylate	Glass ionomer cement
Film thickness	Same or slightly greater	Same or slightly less
Compressive strength	Lower	Greater
Tensile strength	Greater	Lower (GICs tend to be brittle)
Solubility	Lower	Greater
Working time	Comparable	Slightly shorter
Setting time	Comparable	Longer (requires initial isolation)
Bonding (ZNP only bonds mechanically)	Mechanical, some chemical	Mechanical and chemical
Biological reactions	Less	Less but can cause prolonged hypersensitivity reactions

Table 28.1 Approximate composition of zinc phosphate (ZNP) cement

Powder	Liquid		
Zinc oxide (ZnO)	90%	Free phosphoric acid	38%
Magnesium oxide (MgO)	8%	(H ₃ PO ₄)	16%
Silica (SiO ₂)	1.4%	Al and Zn buffered	Rem.
Barium oxide (BaO)	Minor	H ₃ PO ₄	
Barium sulfate (BaSO ₄)	Minor	Water	
Calcium oxide (CaO)	Minor		

Table 28.2 Average properties of inorganic cements

	ZOE	ZNP	ZNC	GIC
Setting time (min)	7–9	5–9	7–9	7–9
Film thickness (μm)	25	25	25	23
Solubility (%)	0.10	0.15	<0.05	0.9
Compressive strength (MPa)	14	100	57–99	150
Tensile strength (MPa)	2	4	3.5–6.5	5
Elastic modulus (GPa)	0.2	11	4–4.8	4.5

ZOE, zinc oxide–eugenol; ZNP, zinc phosphate; ZNC, zinc polyacrylate; GIC, glass ionomer cement

Box 28.1 Types of permanent dental cement

Zinc phosphate
Zinc polycarboxylate
Glass ionomer
Calcium aluminate/glass ionomer
Hybrid (resin-reinforced) ionomer
Resin

Traditionally, permanent dental luting cements were acid–base setting inorganic zinc oxide–based materials (Box 28.1), but the modern trend is greater use of glass ionomers and resin-based materials (Chapters 29 and 30).

28.1 Zinc phosphate

Until the advent of cements with greatly improved properties, zinc phosphate cement (ZNP) was the primary luting agent but is little used now. ZNP is a powder–liquid system (Table 28.1), but its acid content necessitates pulp protection. Setting occurs by ZnO reacting with H_3PO_4 to form a hydrated amorphous $Zn_3(PO_4)_2$ matrix surrounding unreacted ZnO. Setting occurs within 5–7 min, the low initial pH (about 2.1) increasing to pH 4 within 60 min. Strength increases with time, achieving about 90% (100 MPa) of final strength in 24 h, while solubility decreases. After mixing and cement placement within the crown, the restoration should be seated immediately to ensure low film thickness.

28.1.1 Manipulation

Several factors affect ZNP setting behavior:

1 Powder-to-liquid (P/L) ratio: The P/L ratio affects setting time, strength, and film thickness. Higher P/L ratios raise strength but increase film thickness and lower the setting time.

2 Mixing technique: ZNP requires slow incorporation of small amounts of powder into the liquid, ideally mixing on a cool slab for a more fluid mix and longer setting time. Prolonged mixing disrupts the setting mass, reducing cement strength.

3 Liquid contamination: Phosphoric acid is hygroscopic, absorbing moisture on humid days but losing water on dry days. The liquid must be stored in a closed container to maintain cement properties.

28.2 Zinc polyacrylate (polycarboxylate) cement

Zinc polyacrylate (polycarboxylate) cement (ZNC), introduced in the latter half of the 20th century as an alternative to ZPC, sets by ZnO reacting with an aqueous polyacrylic acid solution to form an amorphous gel matrix of zinc polyacrylate surrounding unreacted ZnO. Physical properties are comparable to ZNP (Table 28.2 and Table 28.3). Despite the low initial pH, dentinal tubule penetration of unreacted acid is limited by its large molecular size and ZNC is blander to the pulp than ZNP. The polyacrylic acid may chemically bond to calcium in enamel, enhancing retention. Presently, this cement is used primarily for long-term provisional cementation.

ZnO-based cements that set through an acid–base reaction are sometimes referred to as hydraulic cements.

28.3 Glass ionomer cement (GIC)

GIC is a two-component system available in mechanically mixable and autospensing systems. The powder is a calcium fluoroaluminosilicate ($SiO_2-Al_2O_3-CaF_2-Na_3AlF_6$) glass whereas the liquid is a

mixture of polyalkenoic acids (e.g., polyacrylic, itaconic, and tartaric acids). During setting, the polyacids attack about 20–30% of the glass, releasing Ca^{2+} , Al^{3+} , and F^- ions that react with acid polyanions to form a salt gel matrix stabilized by bound Al^{3+} ions (Figure 28.1). During setting, chemical bonding occurs through ionic interchange with Ca^{2+} and/or PO_4^{3-} ions in hard tissue. Throughout the initial set (over about 3 h), Ca^{2+} ions react with polyacrylate chains, F^- and PO_4^{3-} ions form insoluble salts and complexes, while Na^+ ions form a silica gel; Al^{3+} ions continue to react for at least 48 h. Thus fully set cement is a composite of glass particles surrounded by silica gel in a matrix of polyanions cross-linked by ionic bridges. The gel matrix also contains small particles of silica gel encapsulating fluoride crystallites.

28.3.1 Manipulation and properties

The P/L ratio for polyacid liquid is 1.3–1.35:1 whereas the more fluid water-based systems use a ratio of 3.3–3.4:1. Optimal cement properties require careful mixing, dividing the powder into two equal portions that are mixed in turn. The working time is short (2 min) unless the mixing is done on a cold slab, but this can reduce strength. GICs are moisture sensitive, requiring isolation during initial setting (about 7 min) to prevent moisture ingress. Properties are summarized in Table 28.2 and Table 28.3.

Fewer biological reactions occur with GIC than ZOE, ZNP, and ZNC but prolonged hypersensitivity can arise. GICs may be antimicrobial and, through fluoride release, provide protection against 2° caries.

Bonding to tooth is improved by dentin smear layer removal with acid followed by dilute ferric chloride solution to deposit Fe^{3+} ions, which promote ionic interactions with GIC. Excessive acid pretreatment denudes the surface of calcium.

28.4 New luting agent technology

The latest technology, hybrid acid-based CaAl/glass ionomer, utilizes **nanosstructurally integrating bioceramics (NIB)** and comprises bioactive calcium aluminate and glass ionomer components that are mixed with water. Setting combines a glass ionomer reaction with an acid–base reaction similar to hydraulic cements. The calcium aluminate component confers unique properties, comparable to GICs, notably a low initial pH (~4) that rises to ~8.5 after 3–4 hours. The high pH assures bioactivity by creating apatite on the cement surface when in contact with phosphate-containing solutions together with an excess of Ca^{2+} ions.

The advantage of NIB materials is that calcium aluminate “fixes” the GIC structure, reducing (possibly preventing) the ionomer glass from continuously leaking over time. Postoperative sensitivity is absent.

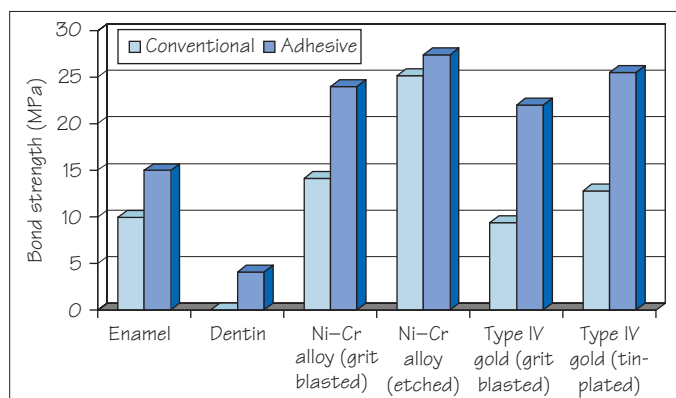


Figure 29.1 Bond strengths for conventional and adhesive resin cements.

Table 29.2 Components of liquid portion of RMGI cements

Component	Function
Methacrylate resin	Permits setting by polymerization
Polyacid	Reacts with the glass causing setting by an acid-base reaction
Hydroxyethylmethacrylate (HEMA)	Facilitates coexistence of resin and acid in aqueous solution and participation in polymerization
Water	Essential for ionization of acid required for acid-base reaction

Table 29.4 Components of resin cements

	Base paste or powder	Catalyst paste or liquid 1	Liquid 2
Single-component paste	Methacrylate monomers + initiator		
Paste-paste	Methacrylate monomers + fillers + initiator	Methacrylate monomers + fillers + activator	
Powder-liquid	PMMA (thickener)	Methacrylate monomers	Catalyst

Table 29.6 Types of resin cements and their applications

Cement type	Clinical application
Light-cure	Metal-free restorations (≤ 1.5 -mm thickness) Nonmetal fixed orthodontic appliances Nonmetal periodontal splints
Dual-cure	Metal-free inlays Metal-free onlays Ceramic crowns Nonmetallic bridges
Self-cure	Metal-based inlays and onlays Ceramometal crowns and bridges Metallic crowns and bridges Metal-based resin-bonded bridges Bonded amalgams Endodontic posts

Box 29.1 Requirements of resin-modified and resin cements

Consistency and handling
Working and setting time
Film thickness
Radiopacity
Strength and wear resistance
Retention (bond strength)
Shade availability
Fluoride release

Table 29.1 Types of RMGI and their applications

Resin-modified glass ionomer	Application
Type I	Luting agents
Type II	Filling materials
Type III	Base/liner
Type IV	Core buildup

Table 29.3 Components of several RMGI cements

Product	Powder-paste	Liquid-paste
A	Fluoroaluminosilicate glass + redox catalyst (encapsulated potassium persulfate + ascorbic acid)	Aqueous solution of methacrylate-modified polycarboxylic acid + HEMA + tartaric acid + activator + water
B	Fluoroaluminosilicate + borosilicate glasses	Complex monomer with carboxylic acid groups + vinyl groups + water
C	Fluoroaluminosilicate glass	Polyacrylic acid + HEMA + tartaric acid + water
D	Fluoroaluminosilicate glass + HEMA + dimethacrylate + initiator	Polyacrylic acid + silica + water + initiator
E	Fluoroaluminosilicate glass + chemical- and/or light-activated initiator	Copolymer of acrylic and maleic acids + HEMA + water + CQ + activator

Table 29.5 Average properties of GICs, RMGIs, and resin cements

	Glass ionomer	Hybrid ionomer	Resin cement
Setting time (minutes)	7	4	4.5
Film thickness (μm)	25	15	10–25
Solubility (%)	0.7	0.2	0.13
Compressive strength (MPa)	150	110	400
Tensile strength (MPa)	5	20	45
Elastic modulus (GPa)	5	5	3.5

Modern dental practice is moving away from hydraulic ZnO-based cements toward **resin-modified** and **resin-based** luting agents. Most of these materials satisfy demanding clinical requirements (Box 29.1) and exhibit excellent adhesion, high strength, biocompatibility, and low solubility.

29.1 Hybrid ionomer cements

The concept of resin-reinforcement of cements was introduced in Chapter 27 (provisional cements). **Resin-modified (hybrid) glass ionomers (RMGIs)** are versatile materials with a variety of applications (Table 29.1); they are available in self-, light-, or dual-cured forms and are supplied as powder–liquid systems, as paste–paste systems, and in automix cartridges and syringes as well as pre-dosed capsules.

The powder component is principally an ion-leachable glass whereas the liquid contains four or more components (Table 29.2), with differences existing between individual commercial products (Table 29.3).

RMGIs may have three setting mechanisms: acid–base reaction, light-activated polymerization, and chemically activated polymerization. Modified composite restorative materials with fluoride release capability (**compomers**, **resinomers**, and **giomers**) are discussed in Chapter 34.

29.1.1 Setting reaction

Initial setting is polymerization of methacrylate groups in the liquid component whereas the slower acid–base reaction provides final strength. Some products use conditioning or bonding agents such as **polyacrylic acid** and **aluminum chloride** or **citric acid** containing **ferric chloride**. Water must be present, either by incorporation or inward diffusion following the initial set, for the acid–base reaction to occur.

29.1.2 Properties

Compared with GICs, RMGIs have improved translucency, the same fluoride release, and greater strength. Bonding to dentin is comparable to that of GICs but RMGIs bond better to resin-based restorations. Initial pH is low (3.5) but increases over time.

Some RMGIs (and GICs), particularly the light-curing types, sorb water and will expand after placement. This hygroscopic expansion can cause fracture of sintered alumina copings of all-ceramic crowns and zirconia crowns.

29.1.3 Applications

RMGIs are used for luting of cast metal and porcelain restorations, posts, and orthodontic appliances, and as adhesive liners for amalgam restorations. They are also used for bases and provisional restorations by changing the P/L ratio.

29.2 Resin cements

Resin cements are essentially low-viscosity composite filling materials having a **resin matrix** containing **silane-treated inorganic fillers**

(silica, glass, or ceramic particles and/or colloidal silica). The polymer systems of resin cements include PMMA or methacrylate copolymers, **bis-GMA** resins, and **urethane dimethacrylate** resins with diluents such as triethylene glycol dimethacrylate (**TEGMA**); see Table 29.4. The fillers, approximately 75% by weight, 47% by volume, have a particle size of approximately 1 μm and provide wear resistance, increased strength, and reduced expansion and contraction.

Polymerization is by conventional chemical-cure or light activation. Dual-cure systems, which utilize both curing mechanisms, are usually radiopaque whereas light-cured systems are not radiopaque but are available in different shades. Most resin cements require a bonding agent to promote adhesion to tooth structure. Some cements are available with high- and low-viscosity catalyst pastes.

29.2.1 Properties

Resin cements have greater strength and lower solubilities than inorganic cements, GICs, and hybrid ionomers (Table 29.5); they form thinner films but set faster. They adhere to hard tissue but bond strengths to tooth and metals are lower than for adhesive resin cements (Figure 29.1).

29.2.2 Applications

A variety of luting materials are available as single-component pastes, powder–liquid systems, and paste–paste systems and are designed for different applications (Table 29.6). Resin cements often are used in conjunction with a dentin-bonding agent (see Chapter 33); metal and ceramic surfaces usually are silanated, while metals can be tin-plated to improve bonding. Unfilled and filled resins are used for bonding of orthodontic brackets.

The disadvantages of resin cements are technique sensitivity, radiolucency with some materials, difficulty in removing excess material, and higher cost.

29.3 Adhesive resin cements

Adhesive resin cements are self-cure systems that incorporate bonding agents to promote adhesion (see Chapter 33). The **dentin-bonding agent (DBA)** provides **coupling** between the resin matrix and the tooth (Figure 33.1).

The 4-META system is a liquid adhesive containing methyl methacrylate monomer and acrylic resin filler and is catalyzed by tributyl borane. Incorporation of polymer beads increases the viscosity to cement consistency.

The cements containing an organophosphate (MDP) are based on bis-GMA (**Bowen's resin**) and silanated quartz filler, provide bonding through reaction of the phosphate end of the chain with calcium of tooth or the metal oxide on metallic restorations.

Adhesive resin cements are sensitive to oxygen and margins must be sealed until set.

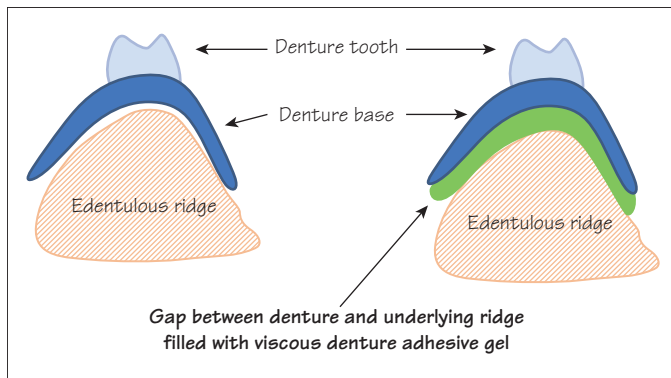


Figure 30.1 Schematic representation of hydration of a denture adhesive to form a gel that swells and fills voids between the CD and its supporting tissues.

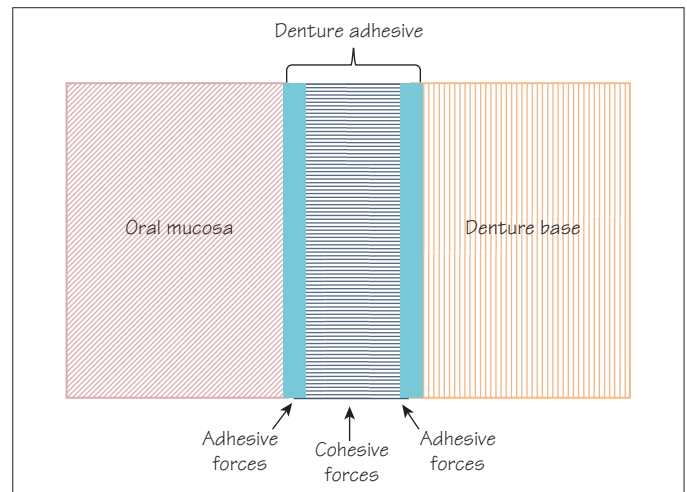


Figure 30.2 Cohesive and adhesive forces in a denture adhesive between a CD base and its supporting structures.



Figure 30.3 Gel denture adhesive applied to the intaglio surface of a lower complete denture.



Figure 30.4 Adhesive strips applied to intaglio surface of a complete upper denture.

Denture adhesives or fixatives are not luting agents per se, but these retention aids can be important in complete denture (CD) therapy by contributing to the “fit,” comfort, stability, and retention of dentures.

Clinically, the space between CD and tissue is filled with saliva, and retention directly relates to the salivary film properties (notably its viscosity and surface tension), the contact area between denture and mucosa, and the wettability of the intaglio surface. Patients with well-fitting dentures and good salivary flow generally experience few problems with stability and retention. Dry mouth (**xerostomic**) patients or those with poorly fitting dentures often resort to denture adhesives to enhance retention, stability, and performance.

30.1 Denture adhesives

Denture adhesives are nontoxic, soluble materials applied as a powder, cream, or pad to the intaglio surface. After application and seating of the denture in the mouth, the adhesive hydrates in the oral fluids to form a gel that swells, filling voids between the CD and its supporting tissues (Figure 30.1). These gels, more cohesive than saliva, adhere to both the intaglio surface and mucosa, functioning as **bioadhesives** and enhancing existing retentive mechanisms. In particular, they optimize interfacial forces and better resist dislodging forces than saliva by increasing the cohesion and viscosity of the fluid layer between denture and basal seat (Figure 30.2).

30.1.1 Dental adhesive materials

Traditionally, denture adhesives were based on **natural gums** (e.g., tragacanth, xanthan, karaya, and acacia), but these had poor cohesion and limited adhesion. Further, these water-soluble gum-based adhesives washed out readily, necessitating frequent replacement.

Modern denture adhesives with superior adhesion and longer service life are based on metal ion salts of two or more polymers. Many contain salts of **carboxymethyl cellulose** (CMC), which hydrates and forms a gel in water, rapidly providing ionic adhesion to both denture and mucosa. The CMC gel increases the salivary film viscosity and cohesion, and eliminates voids between denture and mucosa. These mechanisms enhance interfacial forces between denture and mucosa but they diminish over time due to CMC gel breakdown in the mouth. Consequently, additional components (adhesion “promoters”) are used to augment long-term adhesion.

Most adhesion promoters are hydrophilic colloidal mixed partial salts of lower alkyl vinyl ether–maleic anhydride–type copolymers, notably the inorganic salts of **poly[vinylmethylether maleate]** or “**gantrez acid**.” Typically, these salts are those of calcium, zinc, magnesium, and sodium, although many others can be used. Gantrez acid salts hydrate and form gels that increase adhesion and salivary viscosity but at a slower rate than for CMC. Accordingly, they are longer-acting but slower-onset adhesives although they do undergo cross-linking, which improves cohesion. This effect is pronounced with the binary and triple salts such as calcium–zinc and calcium–zinc–magnesium although zinc-containing formulations now have fallen into disuse. Other polymer systems evaluated as adhesive agents include poly[vinylpyrrolidone] and hydrophobically modified water-soluble polysaccharides, and even enzymatic denture adhesives are being studied.

Additives improve handling and dispensing properties, shelf life, appearance, and taste of denture adhesive formulations. These include petrolatum, mineral oil, and polyethylene oxide as binding agents and

to facilitate placement. Powders often contain silicone dioxide and calcium stearate to minimize clumping. Menthol and peppermint oils are common flavorants, while preservatives such as sodium borate and methyl- or poly-paraben are also present.

Selection of a cream or powder adhesive is based on personal preference. Powders do not provide the same degree of retention or long-term effectiveness as creams but they are used in smaller amounts, are less messy, and are easier to remove. Onset of adhesion appears to be faster for powders than creams. Adhesives with modified chemistry have now become available as thin strips that are placed within the CD. Sorption of saliva causes the strip to swell and initiate the adhesive effect. Adhesive strips are claimed to have the same effectiveness as gels and powders but are more convenient and less messy in use. Figure 30.3 and Figure 30.4 show adhesive gel and strips applied to the intaglio surfaces of upper and lower complete dentures.

30.1.2 Mechanism of retention

Adhesion is through wetting of the denture and mucosa (see Chapters 4 and 5), which allows the denture adhesive to swell and fill voids. Attachment is by mechanical interlocking, i.e. micromechanical adhesion. Other factors such as electrostatic forces between adhesive and substrate microtopography as well as thixotropy also operate.

30.1.3 Clinical use of denture adhesives

Evidence supporting routine use of denture adhesives is limited but they can enhance retention and stability of well-made CDs, particularly for patients with reduced salivary flow. Not all reports wholly support denture adhesive effectiveness, one study suggesting that the only positive effect is a reduction in vertical loosening drops of the distal parts of the denture when the seal was broken.

Denture adhesives are not indicated for improperly fabricated or poorly fitting prostheses. Even when indicated, the least amount of material that is effective should be used, i.e. approximately 0.5–1.5 g per denture.

Because denture adhesives are hydrophilic gels, they provide some cushioning during mastication but they will accrete fluids. Accordingly, daily removal of the adhesive from the intaglio surface and the mucosa is mandatory.

30.1.4 Adverse reactions to denture adhesives

The physical characteristics and retentive capabilities of denture adhesives have been improved by incorporating salts of gantrez acid, typically those based on calcium–zinc and calcium–zinc–magnesium salts, into the formulation. However, leaching of zinc from adhesives can adversely affect the ionic balance within the gut, notably affecting serum copper levels. Quite severe long-term, possibly permanent, neurological problems can develop with excessive use of zinc-containing adhesives, problems exacerbated by near-continuous wearing of adhesive-laden dentures. A further contributory factor is the habit of some denture adhesive users to eat the adhesive because they like the taste! Such behavior clearly would contribute to enhanced zinc ingestion additional to that caused by excessive use of adhesives.

Neurological effects caused by disruption of the zinc/copper balance, however, have been largely eliminated in newer, zinc-free denture adhesives, apparently without adversely affecting denture retention capabilities.

31 Dental amalgam

Conventional amalgam alloy

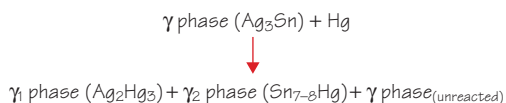


Figure 31.1 Setting reaction of conventional amalgam alloy.

Dispersed-phase (admixed) amalgam alloy

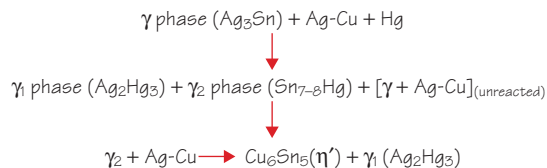


Figure 31.2 Setting reaction of dispersion-modified (admixed) amalgam alloy.

Single-phase amalgam alloy

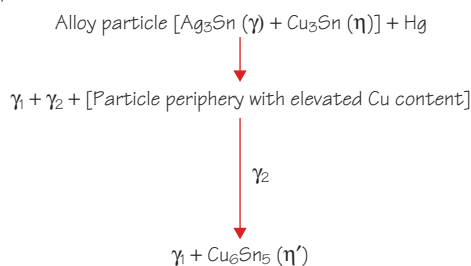


Figure 31.3 Setting reaction of single-phase amalgam alloy.

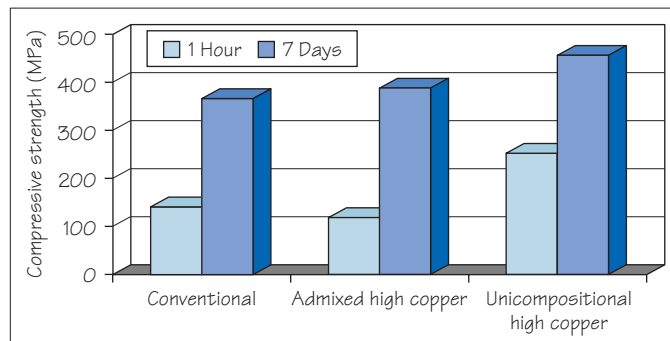


Figure 31.4 Increase in amalgam compressive strength with time.

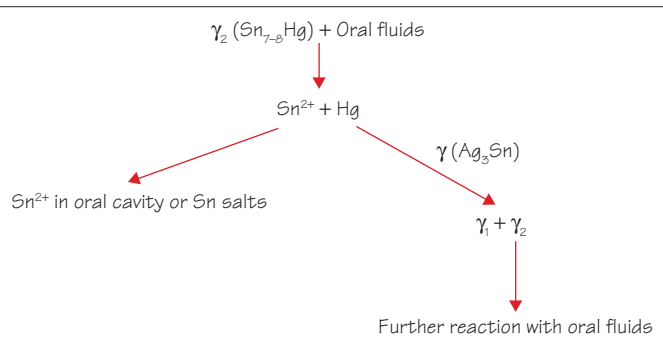


Figure 31.5 Corrosion process with a conventional dental amalgam.

Table 31.1 Specification composition limits (wt.%) of amalgam alloys

	Specification limits	Range	Typical alloy
Silver	40 (minimum)	67–74	69
Tin	32 (maximum)	25–28	25.5
Copper	30 (maximum)	0–15	4.5
Zinc	2 (maximum)	0–2	1

Table 31.3 Average properties of dental amalgams

Product	Tensile strength (7 days, MPa)	Compressive strength (7 days, MPa)	Creep (%)
Conventional (low copper)			
Fine lathe-cut	50	300	6.3
Spherical	55	360	1.5
High copper			
Admixed	45	390	0.45
Unicompositional (spheroidal)	50	450	0.05
Unicompositional (spherical)	55	520	0.09

Table 31.2 Compositions (wt.%) of amalgam alloys

	Conventional (low copper)	High copper (admixed)	High copper (admixed)	High copper (unicompositional)
Particle shape	Lathe-cut or spherical	Lathe-cut	Spherical	Spherical or spheroidal
Silver	63–70	40–70	40–65	40–60
Tin	26–28	26–30	0–30	22–30
Copper	2–5	2–30	20–40	13–30
Zinc	0–2	0–2	0	0
Indium	0	0	0	0–5
Palladium	0	0	0–1	0–1

Dental amalgam is produced by reaction (**trituration**) of mercury with **silver–tin alloy** particles. However, clinical use of amalgam is decreasing due to concern over its mercury content as well as environmental issues associated with waste amalgam disposal.

31.1 Dental amalgam alloy

Alloys are supplied as irregular (**lathe-cut**) particles, **spherical/spheroidal** particles, or a mixture of both, and they consist of silver and tin with lesser contents of copper and zinc (Table 31.1). **Zinc-containing** alloys have $\geq 0.01\%$ zinc content whereas nonzinc alloys have $< 0.01\%$ zinc.

Alloy type and the **alloy-to-mercury ratio** affect the properties of the set amalgam. The mix contains insufficient mercury for complete reaction and set amalgam contains 11–13% unreacted Ag_3Sn particles.

1 Lathe-cut alloys: Particles are irregularly shaped ($60 - 120 \times 10 - 70 \times 10 - 35 \mu\text{m}$), primarily Ag_3Sn (γ phase) with lesser amounts of Cu_3Sn (ϵ) and Cu_6Sn_5 (η') phases.

2 Spherical and spheroidal alloys: Particle diameters are $2 - 43 \mu\text{m}$.

3 Tin in conventional alloys: Tin content is less than 30%; higher Sn content adversely affects amalgam properties.

4 Silver and palladium: Replacement of some Ag by Cu increases Cu_3Sn whereas addition of palladium (Pd) improves strength and corrosion resistance.

5 Copper in conventional alloys: Copper content in conventional alloys is $\leq 5\%$, whereas it is 13–30% in high-copper alloys (Table 31.2).

6 High-copper alloys: Particles may be **uncompositional** spherical or spheroidal, or they may be **dispersion-modified**, i.e. an **admixture** of lathe-cut (40–67%) and spherical (33–60%) particles (Table 31.2).

7 Indium: Addition of indium (10–15%) to the mercury decreases the trituration requirement, increases particle wetting, and lowers mercury vapor release due to the lowered mercury-to-alloy ratio and/or formation of indium oxides at the amalgam surface.

31.2 Setting reaction

1 During trituration, mercury diffuses into the γ phase to form two new phases, γ_1 (Ag_2Hg_3) and γ_2 (Sn_{7-8}Hg), which surround unreacted γ particles (Figure 31.1). The Ag_2Hg_3 (γ_1) phase comprises 54–56 vol.% and the weaker and more corrodible Sn_{7-8}Hg (γ_2) phase constitutes 27–35% of the set mass. During setting, amalgams contract by $\leq 0.05\%$ over the first 30 minutes and then expand by up to 0.04% thereafter. Set amalgams contain variable amounts of voids.

2 Setting of dispersion-modified (admixed) amalgam is a multistage process (Figure 31.2). The first formed γ_2 phase reacts with Ag–Cu particles to form Cu_6Sn_5 and more γ_1 , virtually eliminating all γ_2 in the set mass.

3 Uncompositional high-copper alloys comprise finely distributed Ag_3Sn (γ) and Cu_3Sn (ϵ) phases. Upon trituration, $\gamma_1 + \gamma_2$ form at the alloy particle periphery (akin to conventional alloys), which raises the surface copper content through Ag–Cu formation. Then, like admixed

alloys, Ag–Cu reacts with γ_2 to form Cu_6Sn_5 and more γ_1 (Figure 31.3), eliminating γ_2 in the set mass.

4 Setting of amalgam continues over time, the initially plastic mix stiffening and increasing in strength as more γ_1 and γ_2 form (Figure 31.4).

31.3 Physical properties

Properties of dental amalgams are shown in Table 31.3.

1 Amalgam exhibits greater strengths at higher loading rates due to its viscoelasticity. Compressive strength is almost an order of magnitude greater than tensile strength.

2 Conventional amalgam strength is determined by the relative volume fractions of γ_1 and γ_2 phases and the residual unreacted Ag_3Sn (γ) content.

3 Early strength is greater for finer particle alloys but little difference exists after 1 day. Spherical alloy amalgams are stronger and exhibit less flow and lower setting shrinkage than lathe-cut alloy amalgams.

4 Early amalgam compressive strengths are in the order

Uncompositional > conventional > admixed

5 Dimensional changes on setting are

Conventional > uncompositional > admixed

6 Early tensile strengths are in the order

Uncompositional > conventional > admixed

7 Conventional amalgams are subject to creep; despite the γ_2 phase being very susceptible to creep, creep rate is primarily determined by the volume fraction of the γ_1 phase.

8 High-copper amalgams exhibit less creep and less corrosion than conventional amalgams.

31.4 Manipulation and handling properties

Achieving optimal amalgam restorations requires careful attention to proportioning, mixing, and placement.

1 Overtrituration produces a wet-looking amalgam mass that is difficult to remove from the capsule. Overtrituration of a mix with **low mercury-to-alloy ratio** produces a rapid-setting mass that is difficult to carve.

2 Undertrituration of the amalgam mix and/or delayed condensation into the cavity reduces amalgam mass plasticity, leading to more voids, porosity, and reduced strength.

31.5 Corrosion

The corrosion reaction of a conventional amalgam is indicated in Figure 31.5.

1 Corrosion of γ_2 releases Sn^{2+} ions and Hg, the latter reacting with residual γ particles to form more γ_1 and γ_2 , and with γ_1 to form a mercury-rich γ_1 , the latter weakening the amalgam. The tin ions may be released into the oral cavity or undergo reaction to form salts, typically staining dentin.

2 The principal contributor to amalgam corrosion is the γ_2 phase and since high-copper alloys have minimal γ_2 content, they exhibit virtually no corrosion.

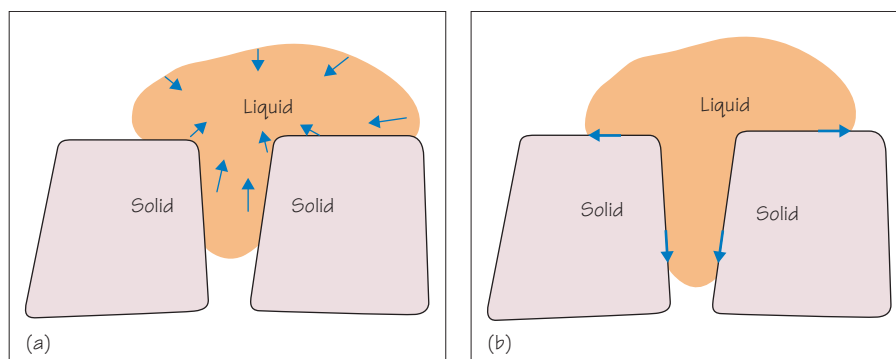


Figure 32.1 (a) Adhesive surface tension exceeds critical surface energy. (b) Critical surface energy greater than surface tension of adhesive.

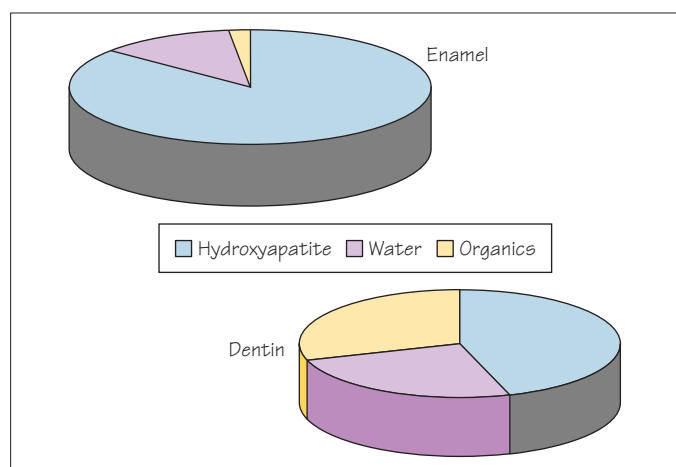


Figure 32.2 Compositions of enamel and dentin.

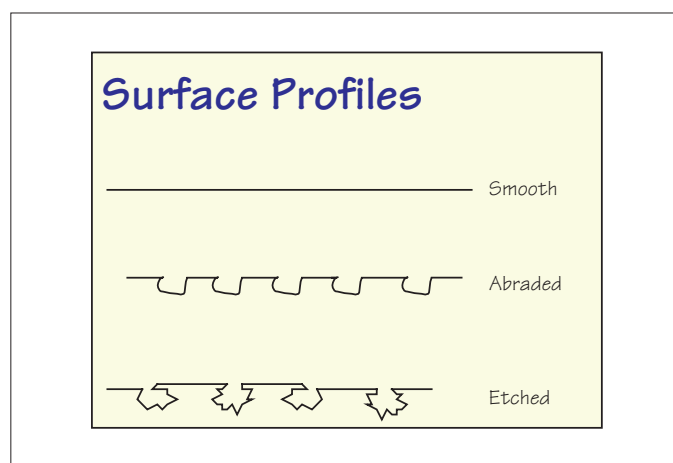


Figure 32.3 Schematic profiles of smooth, abraded, and etched surfaces.

Box 32.1 Factors in dental adhesion

Surface contamination
Surface conditioning
Surface irrigation
Dry/moist field
Adhesive used

Box 32.2 Causes of problems in enamel bonding

Poor surface preparation:

- inadequate etching
 - overetching
 - inadequate or excessive rinsing
- Surface contamination
Improper manipulation of materials
Inadequate adaptation to cavity walls (poor wetting)
Thermal expansion differences
Dimensional change of adhesive on setting

Table 32.1 Common etch patterns of dental enamel

Etch pattern	Characteristic
Type I	Etching of matter between prisms (interprismatic)
Type II	Etching within prisms (intraprismatic)
Type III	Combination of Types I and II

The introduction and proper application of adhesion science within dentistry has changed dental practice, and modern restorative procedures now conserve tooth structure and provide strengthening of the tooth. Adhesive dentistry involves the application and curing of an adhesive resin at the interface between hard tissue and the restorative material. The principles of adhesion and cohesion were discussed in Chapter 4 and Chapter 5.

Adhesive dentistry has three principal steps:

- 1 **Microporosity** in hard tissue is created by acid etching, either by direct application of an etchant or accomplished *in situ* by an etchant/primer/adhesive, as with dentin (Chapter 33).
- 2 A **primer/adhesive** is then applied that wets and penetrates the created microstructure. Since surface energies of etched enamel and etched dentin differ, different primers are required for the two substrates.
- 3 A **resin** is applied to the primed surface so that, when polymerized *in situ*, it micromechanically (and chemically) interlocks with the substrate microporosity.

32.1 Dental adhesion

Because of the nature of the substrate tooth, there is always a large **micromechanical** interlocking (mechanical adhesion) component in adhesive dentistry. Many factors are involved in achieving satisfactory bonding in adhesive dentistry (Box 32.1). Inattention to these reduces the bond strength.

Adhesive spread over the substrate (**wetting**) is dictated by the contact angle (θ°) between the adhesive and the surface (Chapter 5, Figure 5.1 and Figure 5.2). It can be predicted from Young's equation ($\gamma_{SA} - \gamma_{SL} = \gamma_{LA} \cos \theta$) that a high contact angle with poor wetting diminishes adhesion.

The **critical surface energy** (CSE) is the value of γ_{SA} (surface tension between solid and air) when $\cos \theta = 1$; CSE equals the value of γ_{SL} (surface tension between solid and liquid) when the liquid just spreads on the surface. Surface wetting occurs when the liquid surface tension (γ_{SL}) is less than the critical surface energy (Figure 32.1a,b).

The **adhesion quotient** requires the substrate surface energy (γ_{SA}) to exceed the surface tension of the adhesive liquid (γ_{SL}) by 10 dyne/cm. If $\gamma_{SL} \geq \gamma_{SA}$, wetting is poor, adhesion is reduced, and the adhesive tends to pull away from the surface during cure.

32.1.1 Composition of enamel and dentin

Adhesive dentistry is dictated by the composition of tooth substance (see Chapter 6). Dental enamel is highly mineralized while dentin has a high organics content (Figure 32.2). These differences necessitate different approaches to bonding.

32.1.2 Etching of enamel

Enamel bonding requires **conditioning** (i.e., creating **surface topography**) of the relatively inert hydroxyapatite. Abrasion will create rugosity but acid etching, typically with **phosphoric acid** (10–37%

H_3PO_4) gel, introduces microporosity with second- and third-order microstructural facets (Figure 32.3). In particular, etching increases surface area, decalcifies interprismatic material, and creates microspaces and porosities (Figure 6.1). Enamel does not etch uniformly and the commonest etch patterns are indicated in Table 32.1, although other etch patterns have been identified.

Enamel etching varies with tooth mineralization, the location on the tooth, and fluoride levels in the enamel. Because fluorapatite is more acid resistant than nonfluoridated enamel, **fluoridated** and freshly erupted teeth require etching for longer periods than long-time erupted teeth for satisfactory bond strengths. Acid strength does affect etching and bond strength but the effect is less than that of other factors such as surface contamination or fluoridation.

The **ratio of etching time to wash time** is important and generally 15–30 second etches and similar wash times achieve optimal effects. Overetching (≥ 60 seconds) and over-rinsing to remove acid residues are both detrimental to bond strength. Treating etched enamel with **NaF** or **SnF₂** improves bond strength but **APF gel** is detrimental.

32.1.3 Adhesion to enamel

After conditioning, the adhesive/primer is applied to the etched and contaminant-free enamel surface and polymerizes around hydroxyapatite crystals. Adhesive penetration into etched enamel is determined by the **penetration coefficient** (P_C) given by

$$P_C = \frac{\gamma \cos \theta}{2\eta}$$

where γ is the adhesive surface tension, η the viscosity, and θ the contact angle between adhesive and surface. Clearly, lower viscosity and lower surface tension adhesives and low contact angles promote penetration into the enamel microstructure.

Subsequently applied restorative resin will bond to the primer/adhesive layer but the use of combined primer/adhesive systems can be technique sensitive compared with separate primer and adhesive systems. However, **postoperative sensitivity** is reduced with combination systems.

32.1.4 Bonding to enamel

Bonding to enamel is generally well controlled and predictable. Problems such as interfacial gap formation (and subsequent leakage) can arise due to poor cavity design/preparation, clinical manipulation of materials and sometimes to the characteristics of the adhesive system. The most common sources of problems are indicated in Box 32.2.

Enamel bonding has transformed restorative and esthetic dentistry. Further, because of convenience, speed, and many clinical advantages, **direct bonding** of brackets to teeth has almost completely replaced the use of banding with spot-welded appliances in fixed orthodontic therapy.

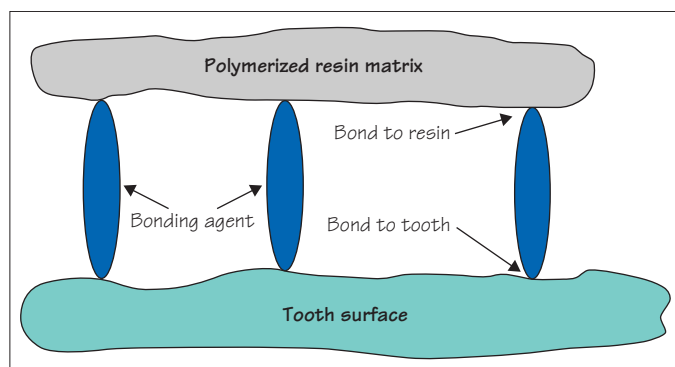


Figure 33.1 Schematic representation of the coupling action of a dentin-bonding agent providing adhesion between the resin matrix and tooth surface.

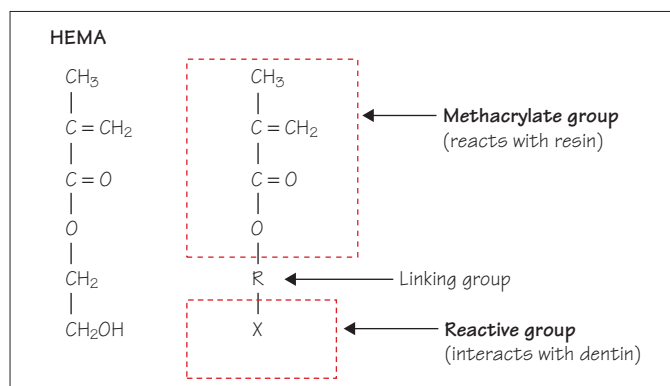


Figure 33.2 Bifunctional HEMA molecule: methacrylate group can bond to resin, whereas the active end can react with dentin substrate.

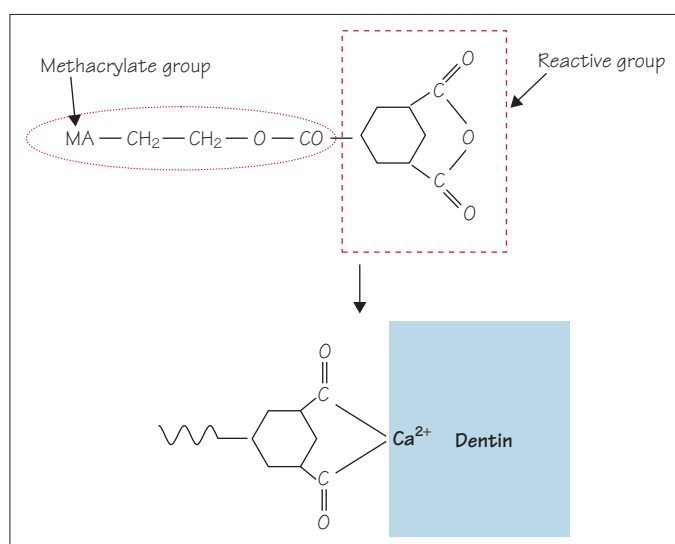


Figure 33.3 Schematic diagram of 4-META and its bond to calcium in hydroxyapatite of conditioned dentin.

Box 33.1 Common dentin conditioning agents

10% phosphoric acid
 35–37% phosphoric acid
 17% ethylene diamine tetraacetic acid (EDTA)
 10% maleic acid
 10% citric acid + 3% FeCl_3

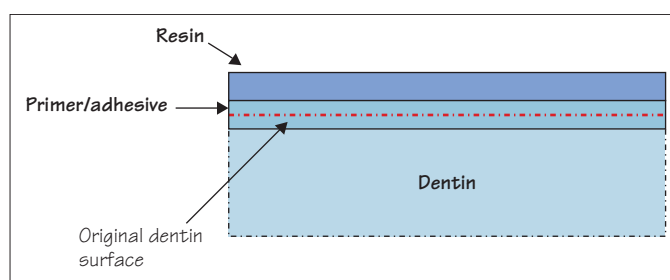


Figure 33.4 Schematic diagram of formation of a hybrid layer on dentin.

Table 33.1 ;Developments (generations) in dentin-bonding systems

Development	Characteristics
First generation	Bifunctional resin bonded to calcium in hydroxyapatite
Second generation	Halophosphorous resin esters bonded to hydroxyapatite
Third generation	Partial smear layer removal, hydrophilic monomers, resin adhesive
Fourth generation	Three steps (conditioning, priming, and bonding) and three solutions
Fifth generation	Separate conditioner; primer and adhesive combined
Single bottle	Conditioner and primer combined; separate adhesive
Self-etching primers	Conditioner, primer, and adhesive combined (single solution)
Sixth generation	No-mix, self-etching systems
Seventh generation	

Table 33.2 Dentin-bonding agents

Bonding agent	Component
HEMA	Hydroxyethyl methacrylate
4-META	4-Methacryloxyethylenetrimellitic anhydride
NPG-GMA	N-Phenylglycine-glycidylmethacrylate
GLUMA	Mixture of glutaraldehyde and hydroxyethyl methacrylate
PENTA	Phosphate penta-acrylate
BPDM	Biphenyl dimethacrylate
NTG-GMA	Reaction product of N(p-tolyl)glycine and glycidyl methacrylate
PMDM	Addition product of pyromellitic dianhydride and 2-hydroxyethyl methacrylate
MDP	10-Methacryloyloxydecamethylene phosphate

Bonding to dentin is more challenging than to enamel because of its more complex structure and composition (Chapter 6), combined with mechanical treatment causing a **smear layer** to form on the surface (Figure 6.2). This smear layer protects the pulp by decreasing dentin permeability but hinders bonding.

33.1 Dentin bonding

Dentin bonding involves a **three-stage process: conditioning, priming, and bonding**, although some systems combine two or more stages into a single step. The term “generation” is used to classify bonding systems. The first three generations, spanning the 1950s to 1970s, were early attempts at dentin bonding with fourth and subsequent generations being developed since the mid-1980s. Acceptable bond strengths are now possible although the later generations represent improvements in methodology and convenience rather than technological advances (Table 33.1).

33.1.1 Conditioning

Access to dentin requires the smear layer to be modified (solubilized) or removed by acidic **conditioners** (Box 33.1). The specific bonding system determines whether or not the smear layer is removed by postconditioning rinsing.

Acid etching of dentin and rinsing removes the smear layer, leaving a smooth surface with patent tubules. Demineralization of the dentin surface and subsurface is known as **the total etch technique**; it leaves a porous surface collagen layer for subsequent bonding by resin infiltration.

For the primer to wet and penetrate moist dentin, it must contain a hydrophilic group and contains a solvent (e.g., acetone) to remove water in the porous dentin surface. Overetched dentin is a poor bonding substrate whereas conditioned dentin must be left moist to prevent collapse of the collagen fibers. The advantage of the total etch technique is that both enamel and dentin can be etched at the same time.

If post-etch rinsing is not performed, the smear layer redeposits on the dentin. This approach is used in systems designed to reduce the overall number of steps in bonding.

33.1.2 Priming

Priming is the **key step** in dentin bonding that promotes interactions between hydrophobic resins and hydrophilic dentin. **Primers (dentin-**

bonding agents, or DBAs) are **bifunctional** molecules, one end being a methacrylate group that bonds to resin and the other a reactive group that reacts with dentin (Table 33.2 and Figure 33.1, Figure 33.2, and Figure 33.3), i.e. primers are **coupling agents**. These bifunctional molecules primarily bond to calcium but may also interact with collagen.

Many bonding agents are available (Table 33.2), and commercial bonding systems commonly incorporate mixtures.

33.1.3 Bonding

The **bonding (adhesive) agent** is a fluid resin that wets and flows over the primed surface, forming an effective bond when cured *in situ*. These polymerizable resins may be two-bottle systems containing an activator (tertiary amine) and a peroxide initiator or one-component systems containing a photoinitiator (camphoroquinone).

33.1.4 Combination systems

To simplify clinical procedures, manufacturers may combine the conditioning, priming, and bonding steps in their systems.

If the primer and conditioner are combined (**self-etching primers**), the smear layer is incorporated within the primer that directly contacts the dentin, the treated surface is not rinsed, and subsequently applied resin bonds to dentin when polymerized. An advantage of self-etching primers is that the dentin is kept moist throughout the bonding procedure although enamel etching is less effective than with phosphoric acid.

Alternatively, the primer and adhesive can be combined and will infiltrate the collagenous network created by conditioning to form a **hybrid (resin infiltrated reinforced) layer** (Figure 33.4). Polymerization of subsequently applied resin bonds everything together.

Sixth-generation (single-bottle) bonding systems use methacrylated phosphoric esters to etch, prime, and bond the dentin in a single step, the methacrylate grouping being able to copolymerize with the bonding agent and resin composite.

33.2 Bonded dentin

Despite high bond strengths to dentin (≥ 20 MPa) and the fact that **bond failures** usually involve **cohesive fracture** of the dentin, these systems are not infallible. They are often technique and material sensitive, and successive treatments may be required for good bonding.

Although high bond strengths suggest good adaptation to the dentin, good bonding and absence of leakage are not synonymous. No system provides consistent leak-free restorations and with all systems, greater leakage occurs in the **cervical direction** than in the occlusal and there is greater leakage in **gingival or cervical areas**. Lower bond strengths are found with deeper dentin than closer to the enamel. Bond strength is improved by application under pressure.

Restoration leakage is reduced with **dentin bonding** compared with an enamel-only bonding or in the absence of bonding. **Reduced postoperative** sensitivity is due to a **barrier effect**, i.e. diminished material transfer into exposed dentin and cementum, and protection may not be necessary for deep cavities.

Dentin-bonding agents are biocompatible, eliciting slight or no reaction; any mild pulpal irritation and/or inflammatory reaction improves with time. The long-term bond durability of dentin-bonding systems is largely unknown.

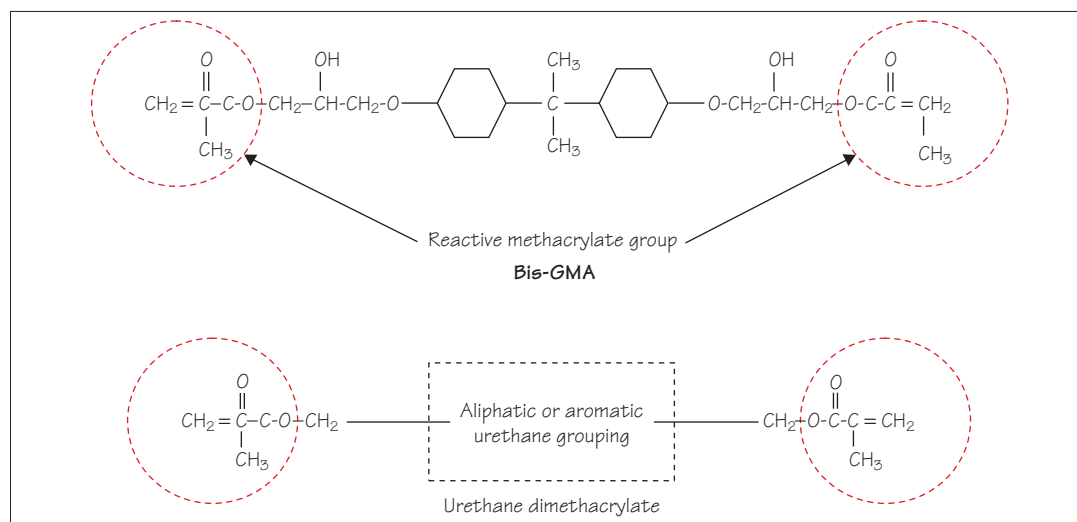


Figure 34.1 Molecular structure of bis-GMA (Bowen's resin) and urethane dimethacrylate (UDMA).

Box 34.1 Clinical requirements of esthetic filling materials

High strength
Low solubility
Comparable elastic modulus to hard tissue
Customizable color
Rapid and controlled setting
Wear/abrasion resistance
Adhesive to tooth material

Box 34.2 Factors determining properties of composites

Monomer/oligomer
Filler type
Filler loading
Curing system
Additives
Shade/color

Table 34.1 Particle size of fillers in composite resins

Type of composite	Filler particle size (μm)
Conventional	5–30
Fine particle	0.5–3.0
Microfine	0.04–0.2
Packable	0.12–0.8
Hybrid	0.02–0.7 (mixture)

Table 34.2 Physical properties of composite restorative materials

Composite	Unfilled	Conventional	Microfilled	Hybrid	Condensable
Filler, vol.%	—	60–65	20–55	60–65	61–70
Filler, wt.%	—	70–80	35–60	75–80	75–85
Compressive strength, MPa	70	250–300	250–350	300–350	200–300
Tensile strength, MPa	24	50–65	30–50	70–90	
Elastic modulus, GPa	2.4	8–15	3–6	7–12	6–12
Thermal expansion, $10^{-6}/^{\circ}\text{C}$	92.8	25–35	50–60	30–40	
Water sorption, mg/cm^2	1.7	0.5–0.7	1.4–1.7	0.5–0.7	
Hardness, KHN	15	55	5–30	50–60	65–75

KHN, Koop hardness number.

Composites are tooth-colored resin-based materials whose properties approach those of tooth substance. They comprise a **resin matrix**, **filler particles**, and a polymerization initiator or catalyst.

34.1 Dental composites

Clinical requirements of composites (Box 34.1) are satisfied through selection of resin matrix, filler type, and filler loading (Box 34.2).

34.1.1 Resin matrix

The first composite materials were based on acrylic resin, which suffered from low strength and high water sorption. Modern composites are based on **dimethacrylate oligomers**, the most widely used being **bis-GMA** and urethane dimethacrylate (**UDMA**), shown in Figure 34.1. These oligomers are viscous liquids, requiring dilution with low molecular weight dimethacrylate monomers such triethylene glycol dimethacrylate (**TEGMA**).

There are some differences in the properties of the resin matrices but the filler has greater influence on the composite properties and most advances derive from modifications in fillers, curing method, and clinical technique rather than resin matrix changes.

34.1.2 Curing systems

Polymerization involves double bonds in reactive centers; composites may be autocuring, light-cured, or dual-cure (both autocuring and light-cured). Surface oxidation of reactive centers forms formaldehyde, whereas air inhibition of the polymerization reaction produces a 25–50 μm thick incompletely polymerized surface layer. Protective gels or overbuilding the restoration and trimming back reduce these effects.

The oligomer shrinks on polymerizing, causing intramolecular and interfacial stresses that are reduced somewhat by water sorption but influence microleakage and marginal adaptation.

1 Autocure (two-component) systems: Once common, autocure systems have largely been displaced by light- and dual-cure systems. These systems require mixing before use but adequate mixing without incorporating air is difficult; **overmixing** reduces working time, with attendant poor cavity adaptation.

2 Light-cured (VLC) composites: VLC composites are single-component systems that have a nearly infinite working time and “set on command.” When irradiated by 470-nm light, the photoinitiator, 0.5 wt.% **camphoroquinone (CQ)**, liberates free radicals. The degree of conversion in light-initiated polymerization is only 60–75%, reduced by poor light access as well as lower light intensity or wavelength variation from curing lamp aging or drops in line voltage. Thick layers of material and/or heavily tinted restorations are more difficult to cure.

3 Dual-cure systems: These are favored where there is restricted light access. After initial light curing, autocuring provides continued polymerization, optimizing the properties of the set material. However, the autocure component contains tertiary amines with an attendant risk of long-term staining from their breakdown.

34.1.3 Filler particles

Fillers are irregularly shaped quartz, borosilicate, barium or strontium glass, and/or organic particles and microfine (fumed) silica that reinforce and modify the resin matrix behavior. Particle irregularity enhances

interlocking with the resin matrix but fillers are **silane-treated** with **coupling agents** such as γ -methacryloxypropyltrimethoxysilane (**GMPTS**) to ensure chemical bonding to the resin. **Conventional** composites use larger particles than modern materials (Table 34.1). Microfine filler particles reduce shrinkage stress.

34.1.4 Filler loading

Filler levels affect composite properties (Table 34.2), with polymerization shrinkage being inversely proportional to filler loading. Higher filler levels increase strength, hardness, and elastic modulus but, above a certain level, abrasion resistance decreases due to the lowered proportion of matrix material and increased elastic modulus increases interfacial stress.

1 Microfilled composites: These have greater thermal expansion and water sorption than conventional composites; strengths are comparable (Table 34.2), but they are easier to polish. Microfills commonly contain pyrogenic (fumed) silica and a composite filler particle, typically a prepolymerized matrix containing pyrogenic silica. They have the lowest filler loading of all composites.

2 Hybrid composites: These contain two types of filler: microfine silica (for wear resistance) and 0.6–1.0 μm ceramic particles for increased strength and reduced expansion/contraction. Total filler content is 75–80 wt.%, with 10–20 wt.% being colloidal silica.

3 Packable (condensable) composites: Introduced as an alternative to amalgam, packable composites have high viscosity permitting comparable packing into cavities. Their fillers differ in shape, composition, and loading from other composites and their matrices are somewhat altered bis-GMA, changes that modify handling properties. They have a greater depth of cure (up to 5 mm) and *in vitro* studies indicate good wear resistance.

34.1.5 Composite strengthening

Larger diameter fillers strengthen composites through particle reinforcement, whereas dispersed submicron particles restrict crack propagation by transferring stress from particle to particle rather than through the matrix. Modern (hybrid) systems use both micron- and submicron-sized particles for strengthening.

Differences in elastic modulus between filler and matrix can cause breakdown due to differential stress effects. The commonest failure mechanism for composites *in vivo* involves matrix breakdown (cohesive failure) or failure at the matrix–filler interface.

34.2 Compomers

Compomers (resinomers and giomers) are polyacid-modified composites containing fluoride-releasing silicate glasses. Designed for lower stress areas in patients at risk of caries, they contain 40–67 vol.% of 0.8- to 5- μm filler particles and set by light-curing with subsequent acid–base reaction upon water uptake. They are similar to, but slightly weaker than, microfills and require DBAs for adhesion.

Formerly single paste materials, newer compomers are two-component powder-liquid or paste-paste systems. Typically, one component comprises a fluorosilicate glass with chemical and/or photoactivated initiators, while the other contains polymerizable methacrylate-polyacid monomers and water. The latter initiates an acid-base reaction immediately upon mixing and renders these materials self-adhesive.

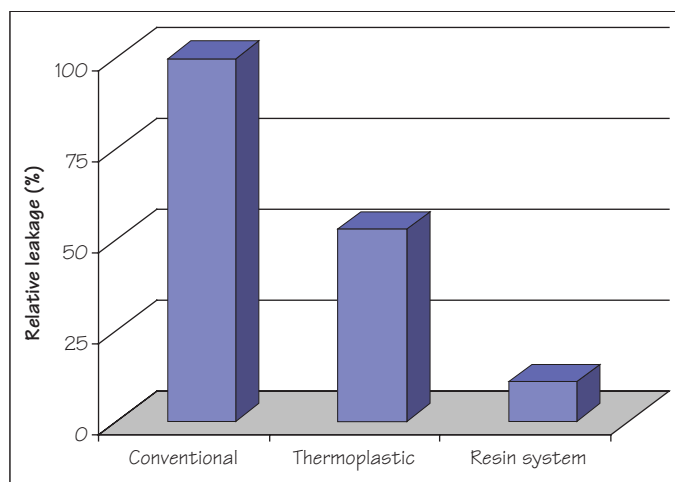


Figure 35.1 Relative leakage behavior of endodontic obturation techniques.

Box 35.1 Factors in successful root canal therapy

- 1 Effective biomechanical instrumentation of the root canal to produce a debris-free surface
- 2 Disinfection and dissolution of organic matter from within the canal to eliminate bacterial pathogens
- 3 Hermetic sealing of the obturated canal

Box 35.2 Ideal characteristics of endodontic irrigants

Nontoxic
 Dissolve/disrupt tissue and debris in the canal
 Low surface tension for good penetration
 Lubricating action on cutting instruments
 Sterilizing/disinfecting action
 Ability to remove smear layer

Box 35.3 Requirements of endodontic sealer cements

Biocompatibility
 Radiopacity equivalent to 3 mm of aluminum
 Resistance to dissolution
 Nonstaining
 Long working time
 Bacteriostatic/bacteriocidal characteristics
 Good adhesion to cementum
 Capability of forming a seal at tooth apex
 Sealing action along canals walls

Table 35.1 Components of traditional endodontic sealer cements

Sealer cement formulation	Powder	Liquid
Grossman's sealer	Zinc oxide Staybellite resin Bismuth subcarbonate Barium sulfate Sodium borate	Eugenol
Rickert's sealer	Zinc oxide Silver Rosin Thymol iodide	Eugenol Canada balsam

Table 35.2 Potential problems with traditional obturation techniques

Cold condensation	Thermoplastic compaction
Filling of large/wide canals	Filling of large/wide canals
Cracking of root dentin during condensation	Overheating of periodontal ligament
Inadequate seal	Material extrusion beyond apex
	Inadequate seal

Over 15% of total U.S. dental expenditures are on root canal therapy. Although treatment has been improved by rotary instrumentation and new obturation materials, absence of pain does not always denote the elimination of continued (or new) periapical pathology.

Three parameters indicate successful endodontic therapy (Box 35.1). These are governed by root canal anatomy, **canal instrumentation**, **smear layer** formation, and canal **obturation**.

35.1 Canal instrumentation

Effective tissue removal, irrigant delivery into the root canal, and smear layer removal are necessary for debris- and pathogen-free canals. Consistent, uniform canal preparation facilitates condensation of the root canal restoration, enhancing the apical seal. Nickel–titanium (**NiTi**) instruments are superior to traditional stainless steel files, producing more uniform and rounder apical canal preparations with less canal transportation. Tungsten carbide and diamond reamers are used to laterally shape coronal access of the canal entrance. Electric handpieces and associated instrumentation are preferred to handheld instruments, achieving faster and more efficient biomechanical canal instrumentation.

Traditionally, metal posts were placed in obturated root canals for subsequent tooth crowning but resin, ceramic, and fiber-reinforced posts are used increasingly.

35.2 Canal irrigation

Irrigants are used to clean and disinfect the instrumented root canal and for smear layer removal; they include organic acids, chelating agents (e.g., **EDTA**), cetrimide, and **hypochlorite** solutions. Optimal irrigant properties are indicated in Box 35.2, characteristics that are necessary because residues within the canal contain infected tissue. Any retained smear layer can seal pathogens within dentinal tubules and preclude adequate sealer penetration and achieving a fluid-tight seal. Ultrasonic agitation, pulsed irrigation, and laser activation of irrigants all facilitate smear layer removal and reduce bacterial counts within canals.

Certain agents, e.g. hypochlorite, are tissue irritants and can cause ulceration of the oral and esophageal mucosa as well as inflammatory reactions if expressed out of the apex into the surrounding bone. Although most irrigants produce virtually debris-free and disinfected canals, some commercial products can leave residues that exacerbate leakage.

35.3 Sealer cements

The prepared root canal is filled with an inert filler such as **gutta-percha** (GP) and a sealer cement, the latter being required because GP does not adhere to tooth. Optimal sealer properties are indicated in Box 35.3. Because sealer cements are placed within a warm moist environment, a long working time and slow setting rate are necessary for placement and radiographic verification of accurate positioning before hardening is complete.

Traditionally, sealer cements were powder–liquid systems based on modified ZOE cements (Table 35.1). A better seal is possible with **AH26**, an epoxy-based sealer, but this material contains formaldehyde—which can induce postoperative sensitivity.

Latterly, photocuring resin sealers have become increasingly popular. These sealers are comparable to dentin-bonding agents and are used in conjunction with resin-based points (cones) and typically are a mixture of hydrophilic difunctional methacrylates. Systems based on these sealers achieve better obturation and sealing of root canals.

35.4 Canal obturation

Customarily, root canal **obturation** (filling) was achieved by cold lateral condensation of GP but canal sealing is improved by thermoplastic compaction techniques. Both approaches achieve satisfactory clinical results but neither technique is without problems (Table 35.2).

Lateral condensation with ultrasonically activated spreaders plasticizes the GP. Plasticization of the GP facilitates condensation and reduces some of the problems cited in Table 35.2. Apical seal is also improved.

Resin-based points and polymeric sealers are used increasingly in endodontic therapy. The filler points are thermoplastic polyester–methacrylate resin blends and may contain bioactive glass filler particles. These resin obturation systems require canal priming/etching agents typically based on HEMA.

There is interest in **mineral trioxide aggregate** (MTA) as an obturation agent because of its strength, chemical stability, potential for bonding to hard tissue, and sealing ability, especially when perforation has occurred. Studies indicate mild inflammation in 17% and 39% of roots with and without an orifice plug, respectively, but no development of severe inflammation. The sealing efficacy of MTA orifice plugs could not be determined.

35.5 Endodontic leakage

Despite the development of newer, more efficient instrumentation and obturation techniques and materials, complete hermetic sealing of root canal apices rarely occurs. Since virtually all endodontic restorations leak to some degree, any apical leakage can lead to endodontic failure. Unfortunately, the ability to sense bacterial breakdown is lost after endodontic therapy. Consequently, patients cannot perceive problems and percolation at restoration margins may be longstanding before being detected. Reinfection also can result from coronal leakage through temporary fillings (see Chapter 36) with significant bacterial leakage through to the apex occurring within 30 days, contributing to endodontic failure.

Root canals prepared with handpiece-driven files and reamers leak less than conventionally prepared canals. Canals obturated with GP and different sealer cements (e.g., ZOE, AH-26, and resins) show comparable leakage behavior although thermoplastic compaction of the GP reduces leakage. Markedly lowered leakage is found with resin obturators and resin sealers (Figure 35.1).



Figure 36.1 (a) Clinical photograph of patient's mouth before treatment. (Courtesy of Heraeus Kulzer US.) (b) Temporized patient with Venus Temp 2 provisional restorations. (Courtesy of Heraeus Kulzer US.)



Figure 36.2 Triad® light-cured provisional restorations. (Courtesy of Dentsply International.)

Table 36.1 Average properties of temporization resins

	Acrylic resin	Bis-Acrylic composite
Polymerization shrinkage (linear %)	4.9	2.9
Transverse strength (MPa)	63	70
Flexural modulus (GPa)	1.5	1.7

Box 36.1 Ideal properties of a provisional restorative material

- Absence of toxicity
- High mechanical strength
- Chemical stability in the oral environment
- Reliable marginal integrity
- Easy handling
- Easy removal
- Good esthetics
- Low cost

Box 36.2 Advantages of light-curing provisional restorative materials

- Fast cure and high curing depth
- Good adhesion to tooth structure
- Low polymerization shrinkage
- Minimal marginal gap formation
- Low water sorption
- Elasticity
- Absence of eugenol
- Removable in one piece
- Absence of stickiness to instruments
- Often available in two or more shades

Provisional (temporary) restorative materials are used for interim sealing of prepared cavities. Provisional restorations are used to maintain esthetics (in the anterior region), to ensure functionality in both anterior and posterior teeth, and to provide protection for preparations as well as close (seal) coronal access openings after endodontic therapy to prevent re-infection before final restoration delivery. The clinical requirements of these provisional materials are indicated in Box 36.1, although no material satisfies all criteria.

36.1 Traditional materials

Until the introduction of zinc phosphate and zinc oxide–eugenol cements late in the 19th century, **gutta-percha** (GP) was the material of choice for interim fillings from about the mid-1800s. GP, composed

of natural gutta-percha, zinc oxide, wax, resin, and metal salts, was softened in a flame and then placed in the cavity. Although now rarely used for provisional restorations, GP is used as a temporary filling for screw-retained implant supported prostheses and as a blocking material between the implant retention screw and occlusal restorations to protect the screw should retrieval become necessary. The disadvantages of GP include low strength, risk of thermal pain, and poor sealing ability, the latter leading to microleakage.

Until recently, zinc phosphate (ZNP) cement was commonly used for permanent luting of cast restorations and as a base (see Chapter 26 and Chapter 28) and has been used as a temporary filling material. ZNP has greater strength and abrasion resistance than ZOE and relatively low solubility in oral fluids. It has poor resistance to masticatory stresses although a higher powder-to-liquid ratio will decrease acidity and increase strength. Claims of pulpal toxicity now appear to be disproved.

ZOE is commonly used as a provisional cement (see Chapter 27) but also as a temporary restorative material, particularly EBA- and resin-reinforced ZOE. Although having anesthetic and antibacterial properties, the eugenol in ZOE can interfere with composite resin polymerization. Various ZOE-based materials are available as OTC products (e.g., Temparin and Dentemp) for emergency sealing of cavities. A widely used P/L provisional restorative is **IRM®**, a polymer-reinforced ZOE material that is superior to other ZOE and ZOE-EBA formulations. Although the eugenol in IRM provides a bacterial barrier, leakage does occur with this material.

Other ZnO-based formulations that do not contain eugenol are known. A popular provisional restorative, particularly for endodontic access openings, is the self-curing (i.e., moisture-curing) **Cavit®**, which contains ZnO, CaSO₄, ZnSO₄, and BaSO₄ as well as ethylene bis(oxyethylene) diacetate and poly(vinyl acetate). Cavit provides a better seal and less leakage than IRM but, in order to ensure optimal strength, thicknesses of at least 4 mm should be used.

Zinc polycarboxylate and conventional glass ionomer cements (GICs) likewise have been used as provisional restorative materials, particularly in endodontics. Bonding to hard tissue is better than with ZnO-based materials but the moisture sensitivity of GICs can be a problem.

A variety of moldable pastes that set through reaction with oral fluids are available. They contain inorganic substances such as ZnO, ZnSO₄, CaSO₄, K₂SO₄, and fluorides as well as glass and/or silica fillers; many also contain methacrylates and urethane dimethacrylates. These resins promote adhesion and often permit light-curing in addition to moisture-curing, i.e. the pastes are dual-cure materials.

36.2 Resin systems

Recently, a number of **single-component light-curing materials** have become available for long-term provisional restorations. These materials have many advantages over traditional materials (Box 36.2); many are radiopaque and incorporate antimicrobial agents. Because such materials are elastic, they are easily placed and readily removed from preparations in one piece without damaging hard tissue. Because of their good adhesion and low sorption, they are suitable for long-term temporaries.

36.3 Coronal restorations

A factor in endodontic treatment failure is coronal **microleakage** since a poorly sealed access opening can allow bacteria and oral fluids to recontaminate the pulp space (Chapter 35). Leaking root canals require

retreatment if left open for longer than 21 days and, ideally, closed with a nonleaking provisional restoration.

Intracanal posts often are used to retain coronal restorations but the seal provided by a cemented post depends on the sealing ability of the luting agent, with adhesive resins and glass ionomers exhibiting less leakage than traditional cements. Further, leakage with resin-coated posts is less than for stainless steel and zirconia because of superior luting agent adhesion to these resin-impregnated posts.

Penetration of microorganisms through provisional restorations can be relatively rapid and has been reported to occur on average within 2 weeks for nonresin materials. Thus, a resin-based temporary restorative material or glass ionomer placed over partially removed composite restorations could result in better resistance to marginal leakage. It appears that glass ionomers provide better coronal seals than bonded composite (or bonded amalgam) with regard to preventing apical bacterial migration, possibly due to better adherence of GICs to the pulpal floor dentin than adhesive resins.

Overall, **core buildup** and/or access closure with adhesive materials provides good long-term coronal leakage resistance, particularly the “sandwich” technique of a glass ionomer base and a composite overlay, possibly because of the protection of the glass ionomer by the composite.

36.4 Temporization

Temporization in the posterior region is commonly achieved using preformed/prefabricated metal (stainless steel, nickel-chromium, tin-silver alloy, or anodized aluminum) or polycarbonate resin crowns. After trimming to adapt the crown to the margins, the provisional prosthesis is luted to the tooth with provisional cement (Chapter 27).

Temporization of anterior teeth, where esthetics is important, usually involves the use of customized restorations that are fabricated using a matrix or an impression made of the prepared teeth (see Figure 27.1 and Figure 27.2). Formerly, acrylic resins were used for this purpose, but now bis-acryl (hybrid acrylic resins) materials are the norm. These self-cure materials have several advantages compared with acrylic resin, including a lower setting exotherm, absence of monomer, less porosity, lower polymerization shrinkage, and greater toughness (Table 36.1). Because most bis-acryl materials are supplied in auto-dispensers, the variation in material properties due to mixing is minimized as is porosity in the cured material.

Not only do bis-acryl materials possess good esthetics and are available in multiple shades, they usually set to a smooth, gloss surface with good color stability and stain resistance (Figure 36.1a, b). The color stability and staining resistance of both classes of material are comparable. A further advantage is that bis-acryl provisionals polymerize rapidly in the mouth, usually within 5-6 min, and they can be modified or repaired with additional bis-acryl or a flowable composite. However, when using these materials, it is advisable to apply a thin film of petroleum jelly to the tooth and any core buildup to prevent any bonding between the bis-acryl and the core material.

Light-cure materials are one-component urethane dimethacrylate resins, e.g., Triad® VLC Provisional Material, often supplied in ropes or as a moldable gel in multiple shades that can be formed to the tooth directly or, more usually, using a matrix or impression as a mold. They can be polymerized in the mouth with a hand-held curing light, although maximal strength requires the use of custom curing units (Figure 22.4). These light-cure materials have sufficient strength to be usable in both the anterior and posterior regions (Figure 36.2).

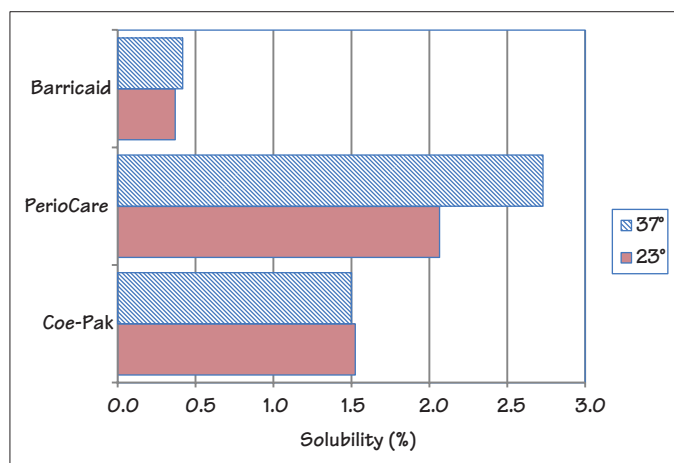


Figure 37.1 Water solubilities of periodontal dressing materials at room and mouth temperature. (Adapted from: J.A. von Fraunhofer and D.C. Argyropoulos, *Properties of periodontal dressings*, *Dental Materials* (1990) 6:51–55.)

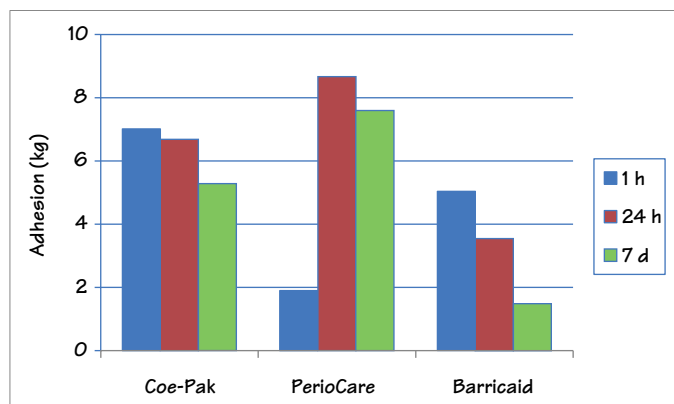


Figure 37.3 Adhesion (separating force) of periodontal dressing materials to hard tissue. (Adapted from: J.A. von Fraunhofer and D.C. Argyropoulos, *Properties of periodontal dressings*, *Dental Materials* (1990) 6:51–55.)

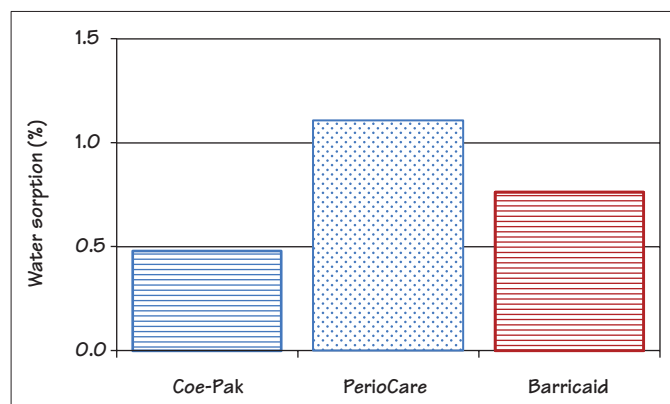


Figure 37.2 Water sorption behavior at 37°C. (Adapted from: J.A. von Fraunhofer and D.C. Argyropoulos, *Properties of periodontal dressings*, *Dental Materials* (1990) 6:51–55.)

Box 37.1 Properties of an ideal periodontal dressing material

- Slow setting
- Smooth, nonirritant surface
- Flexibility
- Good adhesion
- Bacterial growth inhibition
- Dimensional stability
- Nonallergenic
- Resistant to plaque accumulation
- Acceptable taste

Table 37.1 Components of modern periodontal dressing materials

Coe-Pak	Base	Accelerator
	Rosin	Zinc oxide
	Fatty acids	Vegetable oils
	Chlorothymol	Chlorothymol
	Zinc acetate	Magnesium oxide
	Alcohol	Silica
	Cellulose	Synthetic resin
		Coumarin
PerioCare	Paste	Gel
	Zinc oxide	Fatty acids
	Magnesium oxide	Ethyl cellulose
	Calcium hydroxide	Lanolin
	Vegetable oils	Calcium hydroxide
Barricaid	Gel	
	Polyether urethane	
	dimethacrylate resin	
	Silanated silica	
	VLC photo-initiator	

Periodontal surgery involves surgical manipulation of soft tissue to alleviate various problems but, during the histological tissue turnover process, the mucosa requires protection for 6–7 days. In particular, postoperative healing wounds require protection against insult from saliva, trauma, food impaction, and stagnation. Periodontal dressing materials or “packs” are used for this purpose. They also alleviate pain, reduce hemorrhage, facilitate healing, and prevent overgrowth of granulation tissue. The properties of an ideal periodontal dressing are given in Box 37.1.

Traditionally, periodontal dressing materials were eugenol-containing systems derived from ZOE temporary cement, based on the belief that the eugenol had anodyne and antiseptic properties. However, studies indicate that eugenol possesses little antimicrobial activity and, in fact, has toxic side-effects that may delay healing and elicit allergic reactions. Accordingly, noneugenol materials are now used in periodontology; in particular, postsurgical dressing materials are placed in and around surgical sites to provide an obtundent effect and to promote wound healing and epithelial growth. Although these dressing materials are not restorative materials per se, they cannot accurately be described as cements since they are not luting agents.

37.1 Zinc oxide materials

Two ZnO-based periodontal materials are in widespread use: the paste–paste Coe-Pak® system and paste–gel PerioCare®; their principal components are indicated in Table 37.1.

Compared with the amounts of luting agent used for provisional cementation, comparatively large volumes of material have to be mixed for periodontal use and, accordingly, slower setting rates are required. In the case of these two self-curing products, setting occurs within 15–20 minutes. In order that the dressings have a suitable degree of plasticity and reduced brittleness compared with that required for luting or provisional restorations, these materials also have relatively high oil contents. The latter ensures that the mixed material has a smooth texture and is cohesive, so that these dressings are readily formed into ropes to facilitate placement and contouring to promote good adaptation.

The absence of eugenol eliminates the unpleasant taste, odor, and burning sensation found with ZOE materials.

37.2 Polymeric periodontal dressings

A single-component, light-activated periodontal dressing material, Barricaid®, has been introduced for routine periodontal packing and

protection following surgery as well as for use as a protective pack for extraction sites. It is also used to seal in antimicrobial agents placed in periodontal pockets. The components of this gel material are indicated in Table 37.1.

This VLC material gives the clinician total control over the placement and setting of the periodontal dressing and, when cured, the material forms an elastic protective covering over the wound. Further, incremental additions can be made since the material bonds adhesively to itself. Because Barricaid is light-cured, there is no need for mixing and the material is less messy in use and sets “on command.”

Antimicrobials such as antibiotics and chlorhexidine can be incorporated into the unset gel such that these agents are delivered *in situ* with sorption of oral fluids into the cured polymer (see Section 37.3). Such additions, however, do reduce the elastic modulus of the material and its elastic recovery after strain.

37.3 Dressing properties

All three dressing materials are in widespread use and each has certain advantages and drawbacks and, as stated, none contains eugenol. All three materials show some solubility (Figure 37.1). The solubility of PerioCare is greater at mouth temperature than room temperature but this is not the case for Coe-Pak and Barricaid. Likewise, the three materials show sorption behavior at mouth temperature (Figure 37.2). However, whereas water sorption over time actually improves adhesion for PerioCare and has a small but progressive negative effect on Coe-Pak, fluid sorption by Barricaid decreases its already lower adhesion compared with that of the two inorganic materials (Figure 37.3).

Whereas Coe-Pak and PerioCare exhibit good adhesion to hard tissue, their solubility (particularly for PerioCare) suggests that their durability in the mouth may be limited. The polymeric Barricaid shows little solubility but does sorb fluids and this appears to reduce adhesion. Thus, Coe-Pak and PerioCare may be applied directly to the soft and hard tissues, where they appear to anchor well. Barricaid, although easier to manipulate, may require placement such that it encircles the teeth along the arch and extends through embrasures to ensure mechanical locking in position.

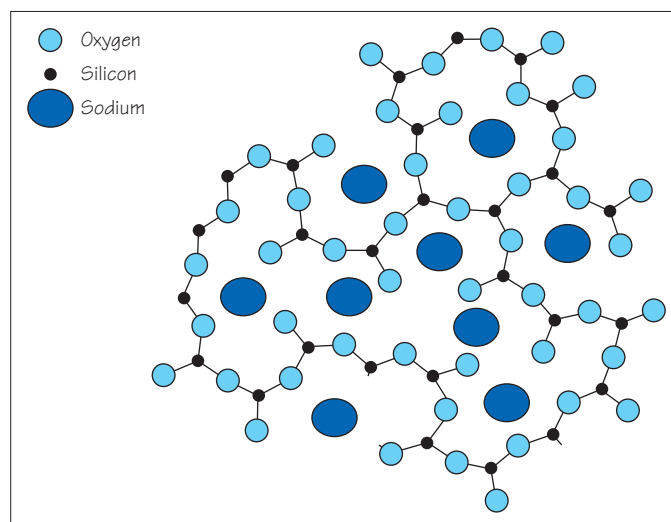


Figure 38.1 Schematic diagram of the two-dimensional structure of sodium silicate glass.

Box 38.1 Clinical applications of dental porcelains

Inlays and onlays
 Jacket crowns
 Veneers
 Castable glass-ceramic crowns
 Denture teeth

Table 38.2 Effects of metal oxides on appearance of porcelain

Metal oxide additive	Coloration effect
Fe_2O_3 or Ni_2O_3	Brown
CuO	Green
TiO_2	Yellow-brown
MnO_2	Lavender
Co_2O_3	Blue
Zr, Ti, or Sn oxides	Opacity
Rare earth oxides	Fluorescence

Table 38.4 Compositional differences between decorative, high-fusing, and low-fusing porcelains

Type of porcelain	Kaolin	Quartz	Binder (feldspar)	Glasses and pigments
Decorative	70	14	15	1
High-fusing dental	4	15	80	1
Low-fusing dental	0	25	60	15

Table 38.1 Compositions (wt.%) of dental and decorative porcelains*

	Dental	Decorative
Feldspar	81	15
Quartz	15	14
Kaolin	4	70
Metallic pigments	<1	1
Appearance	Translucent	Opaque

*Dental and decorative porcelains differ in feldspar and kaolin content.

Table 38.3 Classification of dental porcelains by fusion temperature

Classification	Fusion temperature (°C)
High fusing	1290–1370
Medium fusing	1095–1260
Low fusing	870–1065
Very low fusing	760

All porcelains contain essentially the same components but differences in proportions and firing procedures dictate applications. Dental porcelains, used since 1774, have undergone dramatic changes in composition and properties over the past 15–20 years. Common applications are given in Box 38.1.

38.1 Manufacture

Porcelains comprise crystalline minerals (silica, alumina) dispersed in a glass matrix. The **vitreous (glass) phase** is formed by a **flux** or binder such as **feldspar** when atoms such as sodium disrupt the network of silica tetrahedra to form an amorphous two-dimensional structure (Figure 38.1). The vitreous phase comprises 65% SiO_2 and 15% Al_2O_3 , the remainder being oxides of potassium, sodium, and lithium, as well as boric oxide.

The ingredients (Table 38.1) are blended and fired to a high temperature, forming the vitreous phase. During fusion, feldspar reacts with the outer layers of silica, kaolin, or glass, melding particles together. The molten mass is quenched and ground to a fine powder (the “**frit**”). Fritting may be repeated to incorporate opacifiers, metal oxides and crystalline alumina for coloration or strengthening effects. During porcelain crown fabrication, the fluxes are simply remelted without further reaction between flux and dispersed phase.

38.2 Composition

Dental porcelains contain minimal kaolin and are really **dental glasses**. When ingredients are fluxed with feldspar, the porcelain is known as feldspathic porcelain.

38.2.1 Feldspar

Feldspar—potassium aluminum silicate (orthoclase) and sodium aluminum silicate (albite)—forms the matrix phase; it fuses at about 1290°C , becoming glassy but retaining its form. When mixed with metal oxides and fired, feldspar forms a glass phase that softens and flows slightly (i.e., **minimal slumping**). The powder particles coalesce (fuse without complete melting), a process known as **sintering**. Additions of low-fusing fluxes, e.g. borax, lower the fusing temperature.

Feldspar, used at particle sizes of 1–80 μm , undergoes high firing shrinkage. When heated to $1150\text{--}1530^\circ\text{C}$, feldspar undergoes incongruent melting to form **leucite** (KAlSi_2O_6) in a liquid glass. Leucite has a larger thermal expansion coefficient ($20\text{--}25 \times 10^{-6}/^\circ\text{C}$ compared with $10 \times 10^{-6}/^\circ\text{C}$ for feldspathic glasses), important for fusing porcelain to metal.

38.2.2 Silica

Silica forms an amorphous high melting point glass (fused silica) due to its three-dimensional network of silica tetrahedra. Silica forms a dispersed phase that can bond to feldspar at high temperature.

38.2.3 Kaolin

Low additions of kaolin (china clay, $\text{Al}_2\text{O}_3 \cdot 2\text{SiO}_2 \cdot 2\text{H}_2\text{O}$) confer opacity on porcelain and contribute to matrix formation. It becomes sticky on mixing with water and helps form a workable mass for crown fabrication.

38.2.4 Metallic oxides

Addition of small amounts of metal oxides provides color to porcelains (Table 38.2). They are fused with fine glass and feldspar, the mass being reground and blended with unpigmented frit to achieve the required hue and shade.

38.3 Dental porcelains

Glass modifiers produce porcelains with different firing temperatures, the basis for classifying dental porcelains (Table 38.3 and Table 38.4).

38.3.1 High-fusing porcelains

High-fusing porcelains are primarily used for jacket crowns or denture teeth and consist of previously unfired feldspar, kaolin, and quartz. Firing temperature and time are not critical. They are easily glazed but tend to stain.

38.3.2 Medium-fusing porcelains

Medium-fusing porcelains consist of lower melting frits produced by fluxing with sodium or potassium carbonates and sometimes with CaCO_3 or borax, the additions forming metal oxides on firing. Firing temperature and time are critical and these porcelains are subject to **pyroplastic flow (slumping)**.

38.3.3 Low-fusing porcelains

Low-fusing porcelains are prefused, fritted powder used for metal–ceramic (**PFM**) restorations but are highly modified with fluxes for a narrow fusion range; critical factors are the rate of temperature increase and time at temperature. These porcelains exhibit high pyroplastic flow and a large thermal expansion.

Low-fusing (760°C) porcelains with low leucite content and superior esthetics are claimed to possess superior chemical durability and cause less wear of opposing dentition.

38.3.4 Very low fusing porcelains

Very low fusing porcelains are highly modified prefused, fritted powders and are very useful for low-melting alloys.

38.3.5 Glazes

Glazes are low-fusing, transparent glasses (**veneers**) with thermal expansions matched to porcelain; this avoids **peeling** because of unmatched thermal expansions. Self-glazing of porcelain is commonly used for crowns and bridgework.

38.3.6 Stains

Stains are powdered suspensions of low-fusing colored porcelain in a volatile vehicle applied to porcelain to achieve specific esthetic effects.

38.4 Structure

Porcelains comprise dispersed alumina and quartz in a **glassy matrix**; the higher the firing temperature, the less dispersed phase remains. **Vitreous** glasses are quasi-supercooled liquids. Fluxes modify the matrix by disrupting the network of SiO_4 bonds, lowering the fusion temperature.

Metal ion (Na^+ , K^+ , and Ca^{2+}) disruption of Si–O bonds between SiO_4 tetrahedra forms linear chains of silica tetrahedra with greater mobility at lower temperatures than those within the three-dimensional network. These glass modifiers lower the softening temperature and increase fluidity but large additions reduce chemical resistance; disruption of too many tetrahedra causes **devitrification** (crystallization) of the glass. Boric oxide, used to lower viscosity and the softening temperature, actually forms its own glass network that intersperses with but disrupts and softens the silica network. Alumina does not form a glass itself but incorporates into the glass network, affecting the softening point and viscosity.

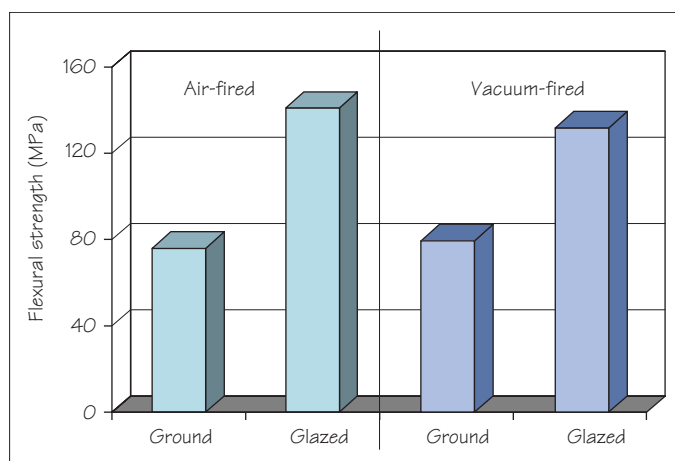


Figure 39.1 Effect of surface condition and firing regimen on porcelain strength.

Box 39.1 Condensation methods for porcelain slurries

Blotting with a brush or 2 × 2 gauze

Mechanically, by vibration or tapping—latter not recommended

Addition of dry powder to surface

Water addition (gravitation method)—not recommended

Table 39.1 Factors in sintering

Sintering factor	Effect
Temperature	Determines glass flow and firing shrinkage.
Glass viscosity	Lower viscosity improves flow of glass during sintering.
Particle size	Range of particle sizes reduces void space; smaller particle sizes increase interparticle contact and more uniform surfaces; affects firing shrinkage.
Surface tension	High surface tension increases flow and bridge formation during sintering.
Air pressure	Increases density by removing air in voids under reduced pressure.

Table 39.2 Stages in firing of porcelain

Firing stage	Effect of firing
Low bisque	Removal of organic matter but minimal shrinkage
Medium bisque	Flux flow and shrinkage
High bisque	Complete vitrification and surface smoothing
Glazing	Glass flow at ceramic surface

Porcelain restorations are fabricated on a die or former that was produced from an impression and cast of the patient's dentition. An aqueous slurry of porcelain powder is applied to the die and then fired (**sintered**) followed by customizing to meet specific esthetic requirements.

39.1 Manipulation

39.1.1 Paste

A slurry is made by mixing porcelain powder with water, starch, sugar or glycol, and commercial additives, the latter increasing viscosity and wettability. The manipulative characteristics of the slurry are determined by porcelain particle size and distribution as well as any organic binders in the liquid. The slurry then is poured into a mold or applied by brush or spatula to a die; the latter process is known as **stacking**.

39.1.2 Condensation

Water removal from the slurry (condensation) is critical for close packing of powder particles and minimizing void space within the mass. Successful condensation reduces (slightly) firing shrinkage, pyroplastic flow, and cracking of fired porcelain while increasing strength. Several methods of condensation are used (Box 39.1), the least effective being water addition because it can wash out finer powder particles while increasing shrinkage by creating more voids in the dried mass. Likewise, tapping the applied mass is not recommended because of possible disruption and forced settling of the condensing mass.

39.1.3 Dies

A **platinum matrix** or high-temperature die material is used as a substrate for building (stacking) porcelain when preparing ceramic restorations. The platinum matrix must be carefully adapted to the die using folded (tinner's) joints.

39.1.4 Drying

Stacked porcelain must be dried slowly (near an open oven) to remove water. This prevents porosity, steam generation, and/or cracking of the porcelain.

39.1.5 Firing (Sintering)

Sintering entails formation of **glass bridges** between unfused porcelain particles, several factors being involved (Table 39.1). The type of porcelain determines sintering temperature, e.g. lower temperature glass flow with low-fusing porcelains than with higher fusing material. The sintering temperature also determines the porcelain firing shrinkage due to better flow from decreased glass viscosity at higher temperatures. Likewise, higher surface tension improves flow and bridge formation although this decreases at higher temperatures.

Firing under reduced pressure (ca. 50 mm Hg) removes air within voids but "vacuum" fired porcelain requires special pigmentation to avoid breakdown under reduced pressure.

The stages in firing are given in Table 39.2. Several firings are performed to different temperatures, allowing several layers of porcelain to be applied.

1 Glazing: Porcelain is glazed by (i) rapid high bisque firing with minimal high temperature "hold" or glaze application (overglazing) or (ii) smoothing by flow of highly modified (fluxed) glass. Glazing improves porcelain strength (Figure 39.1). Porcelains are rarely annealed.

2 Firing shrinkage: Firing shrinkage, usually 15–35%, depends on firing temperature. Shrinkage of a properly condensed mass generally is independent of condensation method, being determined by the particle size and distribution. Slow cooling from the firing temperature reduces superficial thermal stress crack formation in the porcelain surface.

39.2 Physical properties

39.2.1 Strength

Porcelains and glasses are hard, brittle materials; their strength is determined by surface irregularities, internal voids, and porosities. Cracks are the cause of brittleness, commonly initiating from the interior core layer and propagating outward toward the surface. Fatigue strength is limited by crack propagation and thermal stress cracks; scratches from surface grinding can act as crack initiators (**stress risers**). Glazing inhibits crack propagation.

Porcelain susceptibility to failure is exacerbated by its propensity for undergoing the time-dependent process of **static fatigue**. This causes reduced strength even in the absence of external loading although failure is accelerated under dynamic mechanical loading. The precise causes of static fatigue are not known but are probably related to hydrolytic reactions involving the internal structural components of porcelain.

Total elongation is less than 0.1% and bending by 0.01% causes fracture. Porcelains are much weaker under tensile or transverse loading than in compression. Transverse strength can be increased through refinements in formulation and processing, e.g. "high-strength" porcelains (Chapter 40).

Porcelains are virtually chemically inert and are highly resistant to chemical attack but physical property considerations can limit porcelain use in posterior regions.

39.2.2 Porosity

Entrapped gases during **vitrification** cause porosity but can be removed by (1) firing under vacuum, (2) firing in selective atmospheres, or (3) cooling under pressure. Strength increases from reduced porosity are limited to 25% for impact strength and 18–30% in compressive strength. Although reduced porosity improves esthetics, it has little effect on tensile strength.

39.2.3 Hardness

The hardness and abrasion resistance of porcelains make them difficult to polish although a polished surface is less resistant to plaque accumulation than a glazed surface. Whereas porcelains can abrade opposing tooth enamel, such abrasivity varies with the ceramic characteristics and is often unpredictable since it is related to the material's internal microstructure as well as the nature of the opposing dentition.

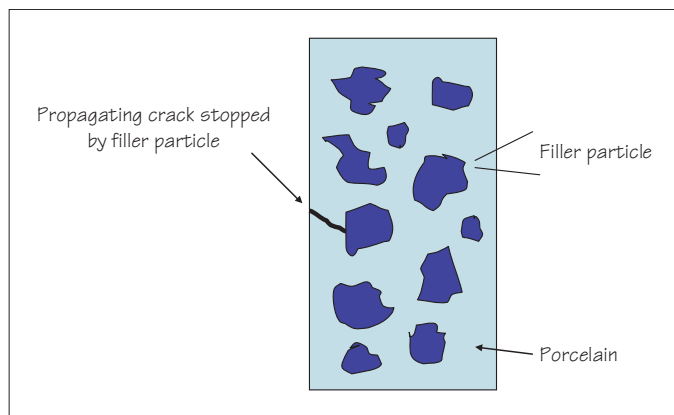


Figure 40.1 Reinforcing filler particle acting as a “crack stopper” for a propagating crack.

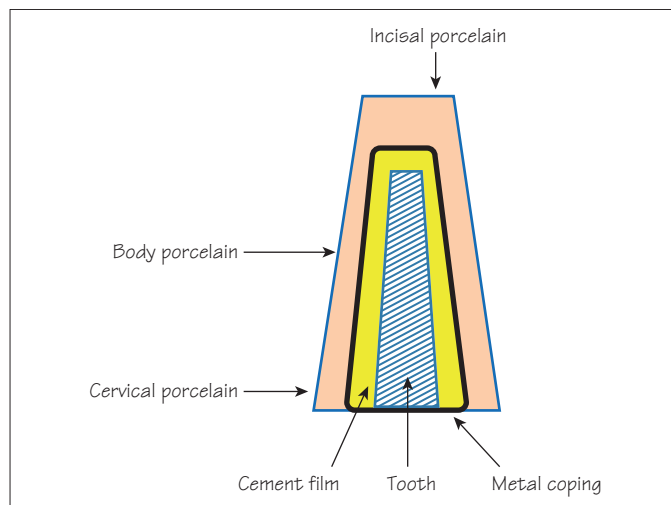


Figure 40.2 Schematic representation of a PFM (metal-ceramic) crown.

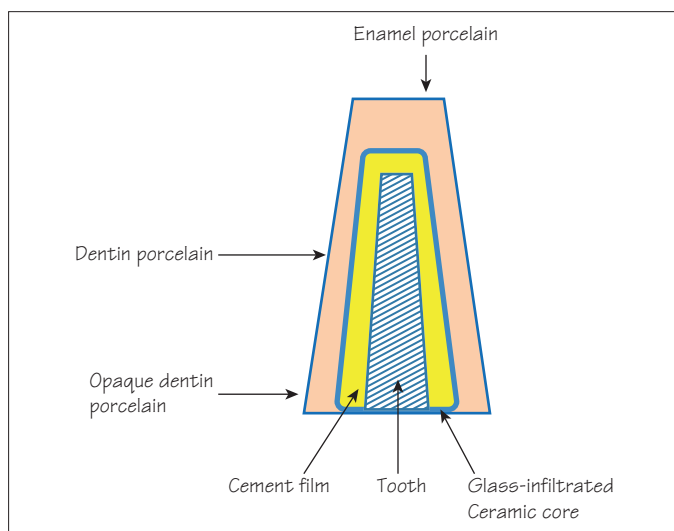


Figure 40.3 Schematic representation of a crown constructed with a glass-infiltrated (In-Ceram) core.

Table 40.1 Strengths of dental hard tissues and dental ceramics

Material	Flexural strength (MPa)	Fracture toughness (MN/m ^{3/2})
Enamel	65–75	1
Dentin	16–20	2.5
Feldspathic porcelain	60–110	1.1
Leucite porcelain	120–180	1.3
Aluminous porcelain	130–140	2.0
Glass-infiltrated alumina	400–600	4–6
Spinel-infiltrated alumina	325–410	2.4

Porcelain restorations can be fabricated in a range of shades and translucencies that, with esthetics and chemical resistance, make them almost ideal replacements for lost hard tissue. Further, being a good thermal insulator, porcelain protects the pulp against heat insult even with marked hard tissue loss.

Limitations of porcelain were indicated in Chapter 39. Despite improvements that are possible by controlling manipulative and processing variables, there are strength limitations.

40.1 Porcelain strengthening

Porcelain is strengthened by inhibiting crack formation and propagation at the surface and one approach is **ion strengthening**. This technique involves replacing sodium ions in the surface of the fired porcelain by bulkier potassium ions, thereby creating compressive stresses at the surface and eliminating the tensile stresses leading to crack propagation. Ion strengthening involves coating the fired component with an ion-exchange paste and then heat-treating to cause the sodium–potassium exchange.

It is possible to eliminate surface stresses through a heat-treatment regimen, typically during initial firing. Since the inner layers cool more slowly than the outer layers, the latter are placed in compression. This approach is reasonably effective but involves careful attention to firing procedures.

40.1.1 Aluminous porcelain

Porcelain can be strengthened by incorporating up to 40% alumina particles. These act as “crack stoppers” that prevent crack propagation, either directly or by extending the crack path length (Figure 40.1). Aluminous porcelains have a 50–100% greater flexural strength than feldspathic porcelains but have comparable thermal expansions and elastic moduli.

Because of high opacity, aluminous porcelain is used as a core material covered with a veneer. However, large amounts of tooth may need to be removed during preparation to accommodate the bulky restoration. Further, these porcelains are prone to strength degradation if they contain porosity.

40.1.2 Alumina inserts and cores

Inserting small sheets of alumina palatally in a crown also increases porcelain strength. Greater strengthening is possible by constructing the crown on a hard, opaque sintered alumina core that is less susceptible to crack propagation. Several commercial systems are available for producing such cores.

The In-Ceram process involves painting a water-based alumina powder slip onto a plaster die and the plaster absorbs the water, a 0.5 mm thick alumina layer remains on the die surface. When fired for about 2 hours at 1200°C, the die shrinks allowing easy removal of the alumina core, the outer surface of which is painted with a lanthanum glass powder slurry. Firing at about 1100°C allows the glass to flow and infiltrate the core surface, achieving strengthening and improved esthetics. After surface blasting to remove excess glass and smooth

the surface, the core is refired at about 960°C to complete the infiltration process. The resulting sintered alumina core ceramic has a flexural strength some three times greater than that of aluminous porcelain and four to five times greater strength than unreinforced porcelain.

Alumina cores are also fabricated in the CeraOne and Procera systems. These systems essentially press alumina onto a metal die. The pressed shape is removed from the die and sintered. These cores, however, have no glassy phase between the particles but achieve good clinical results.

Finally, feldspathic porcelain dentin and enamel layers are built up on the alumina core in the standard manner to complete the restoration.

40.1.3 Spinel-based cores

Magnesium aluminate spinel-based cores have been used in place of pure alumina; the spinel is formed by interaction of magnesium oxide with the alumina to form a mixed metal oxide or spinel. These cores have superior translucency to sintered alumina cores but somewhat lower flexural strengths.

Greater flexural strength cores (800 MPa compared with 400–600 MPa) are possible by incorporating zirconium oxide (zirconia) within the alumina slip powder. However, sintered alumina–zirconia cores are relatively opaque, which is difficult to mask.

40.1.4 Leucite-reinforced porcelain

The presence of up to 45 vol.% of leucite strengthens feldspathic porcelain because the high leucite content elevates the thermal expansion coefficient, generating compressive stresses in the glassy phase during cooling. This increases flexural and compressive strength whereas the finely dispersed leucite grains modify abrasion behavior so that wear rates are similar to enamel.

These porcelains have good esthetics and are used for porcelain-fused-to-metal (PFM) restorations because of comparable metal–ceramic thermal expansions. A pressed leucite reinforced ceramic system, IPS Empress, incorporates fine leucite particles to increase flexural strength through the increased volume of the particles.

Porcelain and hard tissue strengths are shown in Table 40.1.

40.1.5 Metal-bonded porcelain

Many problems with porcelain are minimized by the ceramic–metal or porcelain-fused-to-metal (PFM) restoration, which takes advantage of the support provided by a metal substructure provided there is good bonding between the metal coping and porcelain. After casting a metal alloy to the restoration shape, high-leucite porcelain is fired onto it with the metal providing support for brittle porcelain. Strong bonding between porcelain and metal is possible with many metals (see Chapter 17). Disadvantages of PFM restorations include radiopacity, lack of natural aesthetics, and, sometimes, metal biocompatibility questions.

The differences between a PFM (metallo-ceramic) crown and an In-Ceram crown with a glass-infiltrated ceramic core are shown schematically in Figure 40.2 and Figure 40.3.

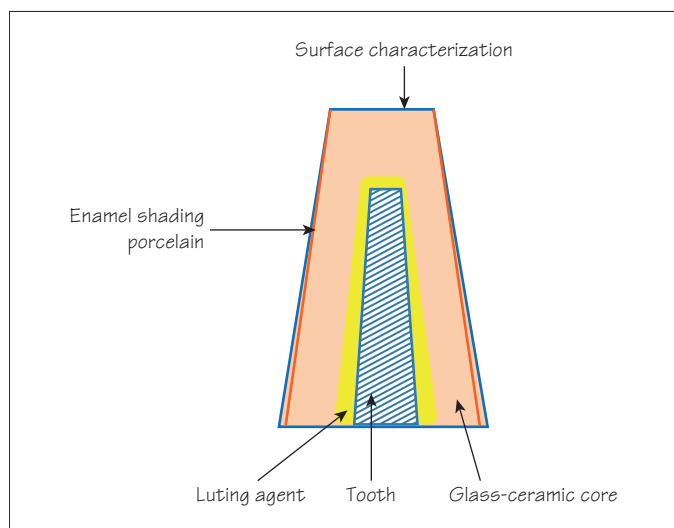


Figure 41.1 Schematic representation of a crown constructed with a glass-ceramic (Dicor) core.

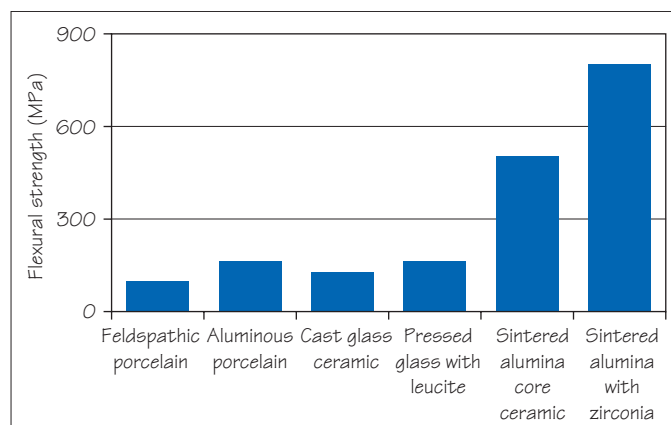


Figure 41.2 Flexural strengths (MPa) of porcelain and castable glasses.

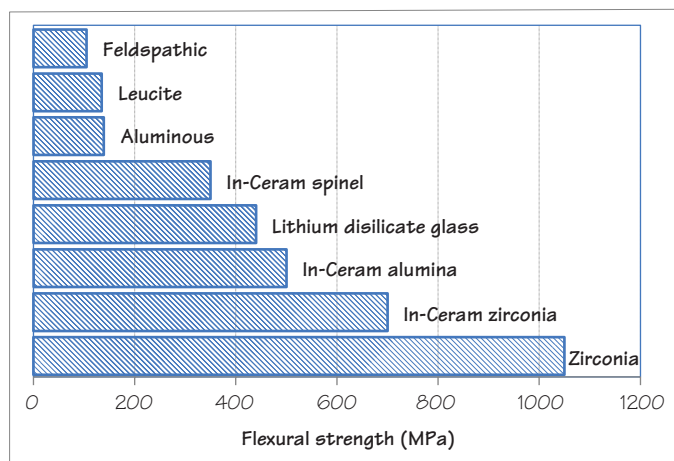


Figure 41.3 Average flexural strengths (MPa) of porcelains and advanced ceramic systems.

Table 41.1 Dental ceramics and methods used for restoration fabrication

Fabrication method	Crystalline phase
Sintered	Alumina Fluorapatite Leucite
Hot-pressed	Leucite Lithium disilicate Lithium phosphate
Slip-cast	Alumina Magnesium–alumina spinel Zirconia–alumina spinel
Machined	Alumina Feldspathic porcelain Fluoromica Leucite Zirconia (Y-TZP and Ce-TZP)

In recent years, esthetic considerations regarding metal–ceramic systems and questions about their reliability have stimulated a growing trend for all-ceramic restorations (ACRs). Originally ACRs were based on porcelain reinforced with alumina cores and leucite (Chapter 40) but newer, far stronger ceramics containing up to 99 vol.% of a **reinforcing crystalline phase** are now available (Table 41.1). Techniques such as sintering, hot pressing, slip-casting, and machining are used to fabricate restorations and there is increasing use of high-strength castable cores with the restoration body and incisal region of conventional porcelain for esthetics.

41.1 Hot-pressed ceramics

Various ceramics can be sintered and shaped by external pressure at high temperature (**hot pressing** or **high-temperature injection molding**) to form all-ceramic restorations, the advantages being good dispersion of the crystalline phase, reduced porosity, high density, and excellent strength.

41.1.1 High-leucite molded glass

High-leucite (35–55%) fine-particle glass is melted at 1150–1180°C and hot pressed into a refractory mold prepared by the lost wax process. The coping is ground to shape followed by application of an outer layer of incisal porcelain (i.e., a porcelain veneer) and conventional firing. Restorations have high strength but equipment cost is high and processing is complex.

41.1.2 Lithium ceramics

Lithium disilicate, quartz, and cristobalite form a hot-pressable glass–ceramic with greater strength than most other all-ceramic systems. The thermal expansion is compatible with that of low-leucite porcelain.

Lithium phosphate in a glassy matrix is a ceramic that is hot-pressable onto zirconia endodontic posts.

41.1.3 Slip-cast ceramics

Slip-casting is the condensation of an aqueous porcelain slip or slurry onto a refractory die; it is fired and then infiltrated with molten glass to produce a low-porosity restoration with reduced defects and high toughness. However, the fatigue resistance of these restorations has been questioned. Slip-casting was mentioned in Chapter 40 but is mentioned again here for completeness:

1 Alumina slips: Slips containing >90% 0.5–3.5 μm alumina are fired, infiltrated with lanthanum glass, refired, and then veneered.

2 Spinel: **Magnesium spinel** (MgAl_2O_4) and **zirconia–alumina spinel** also are being used in slip-casting. Restorations have high strengths and lower opacity than other ceramics.

Ceramics processed by slip-casting are reported to have a 30–35% greater flexural strength than their machinable counterparts although there appears to be no difference in fracture toughness.

The advantage of the slip-cast technique for core fabrication is that there is limited shrinkage although the porosity of slip-cast cores (8–11%) is greater than that of sintered stabilized zirconia (see Section 41.2). This accounts for the generally lower mechanical properties of slip-cast compared with sintered ceramics.

41.1.4 Castable glass–ceramic

Glass–ceramic of fluoromica crystals is cast in a refractory mold and then heat treated. This induces transformation of glass (clear) to ceramic (translucent), during which 1 μm mica platelets are formed. The crystal nucleation and growth process is known as **ceramming**, and the tetrasilicic fluoromica platelets comprise 55 vol% of the glass–ceramic. Ceramming increases the strength and toughness, abrasion resistance, and chemical stability of the material as well as decreasing translucency. The surface may be stained and glazed, but restoration translucency necessitates shaded cements. A schematic representation of a Dicor crown is shown in Figure 41.1, indicating the glass–ceramic core compared with the cores in PFM and In-Ceram crowns (Figure 40.2 and Figure 40.3). Despite acceptable esthetics, Dicor is no longer available, possibly because low (tensile) strength limited its use to lower stress areas due to susceptibility to fracture.

Flexural strengths of porcelains and cast glasses are shown in Figure 41.2.

41.2 Machinable ceramics

Increasingly, restorations now are milled from monolithic blocks of “zirconia” using **CAD-CAM technology** (see Chapter 42). Zirconia (zirconium oxide, ZrO_2) occurs naturally as baddeleyite or zircon (basically $\text{ZrO}_2\cdot\text{SiO}_2$, containing 80–90% ZrO_2) but not as the pure oxide. Zirconia has three crystal structures: monoclinic at room temperature, tetragonal at $\sim 1200^\circ\text{C}$, and cubic at 2370°C , transferring from one crystalline state to another during firing. The tetragonal form is stabilized at room temperature by adding 3–8 wt.% of yttria (Y_2O_3) or ceria (CeO_2). Yttria-stabilized zirconia is referred to as Y-TZP (yttria-stabilized tetragonal zirconia polycrystals) and ceria-stabilized zirconia as Ce-TZP. Ce-TZP has better thermal stability and resistance to low-temperature degradation than Y-TZP under similar conditions. The stabilized forms of zirconia are used in CAD-CAM technology.

The clinical usefulness of stabilized zirconia is that, although the tetragonal form is stable at room temperature, it is metastable in that trapped energy within the material can cause it to revert to the monoclinic state with a corresponding 3–5% volume increase. If cracking occurs, the highly localized stress ahead of the propagating crack will trigger crystallographic transformation in the vicinity of the crack tip such that the associated volume increase, and change of tensile to compressive stress, will close the crack. In other words, crystalline transformation decreases the local stress intensity such that stabilized zirconia has high flexural strength (ca. 900–1200 MPa) and excellent resistance to stress fatigue.

The applications and flexural strengths of porcelains and high-strength ceramics are summarized in Table 41.1 and Figure 41.3.

41.3 Veneers

After removal of 0.5 mm of labial enamel, **veneers** (0.5–0.8 mm thick) are fabricated from porcelain, glass–ceramic, hot-pressed ceramic, and even by CAD-CAM techniques. The veneers are bonded to the prepared enamel with dual-cure cement and markedly improve the appearance of stained or discolored teeth.



Figure 42.1 Dental laboratory CAD-CAM systems. (Courtesy of Sirona Dental Inc., Charlotte, NC.)



Figure 42.2 E4D CAD-CAM system for the dental office. (Courtesy of D4D Technologies, Richardson, TX.)



Figure 42.3a Patient before restoration. (Courtesy of D4D Technologies, Richardson, TX.)



Figure 42.3b Patient after restoration. (Courtesy of D4D Technologies, Richardson, TX.)



Figure 42.4 Finishing and characterization of a CAD-CAM ceramic restoration. (Courtesy of Sirona Dental Inc., Charlotte, NC.)

CAD-CAM is the abbreviation for computer-aided design–computer-aided manufacture, a long-established technology in industry and of growing importance within dentistry, potentially enabling teeth to be restored in a single appointment. A number of commercial systems are available, including the E4D, CEREC (CERamic REConstruction), 3Shape Dental, and Cera systems. Technological advances include versatile software, direct digital recording of the dentition, and greater speed and accuracy in milling operations. Laboratory and chairside CAD-CAM systems are shown in Figure 42.1 and Figure 42.2.

42.1 Digital imaging

Although specific details vary with each system, the same basics underlie digital restoration processing, with everything starting from a three-dimensional image used by computer software to design the restoration. For this, undercuts are eliminated or blocked out before the tooth is sprayed with a thin layer of blue antireflective contrast medium and recorded by a 3D imaging camera. Imaging systems also can record images of the adjacent and opposing dentition as well as bite registration data. After information is uploaded into the computer, a data file is assembled that, together with the computer's internal library of tooth shapes, is used to design the restoration.

After restoration design, the data file is fed to an in-office system or to a remote laboratory where the restoration is milled from a monolithic (solid) block of ceramic or composite (Figure 42.1 and Figure 42.2). After try-in, the restoration can be adjusted to ensure an exact fit with the correct occlusion. Milling digital restorations from a monolithic block enables better blending with the surrounding dentition for a more esthetic outcome. Ideally, the restored tooth will have an anatomically and functionally perfect restoration (Figure 42.3a, Figure 42.3b).

42.2 Restorative materials

Digital restorations are predominantly fabricated from high-strength ceramics (Chapter 41). Additionally, high-strength composite materials derived from restorative resins have been developed specifically for CAD-CAM applications.

Ceramic digital restorations have high strength, natural translucency and fluorescence, good fit, and in many cases, e.g. the leucite-reinforced

glass–ceramics, they can be polished and characterized (Figure 42.4). They have remarkable chemical resistance and problems such as plaque buildup and staining found with other materials are absent. Not all materials, however, possess high flexural strength or fracture toughness and they can cause wear of opposing dentition.

The CAD-CAM composite resins do have high flexure strength and fracture toughness and their wear characteristics are comparable to those of enamel, obviating damage to opposing dentition. They are easier to finish and polish than ceramics and can be easily characterized using light-cured composite stains, and restorations can be repaired in the mouth. However, like all resin-based restorations, they can be subject to staining and wear in use.

Because composite restorations have a resin matrix, treatment of the interior (fitting) surfaces to facilitate bonding is relatively easy. In contrast, ceramic restorations may require hydrofluoric acid etching and silane treatment prior to adhesive bonding.

Being machined from monolithic material, restorations are generally stronger than those that are incrementally constructed, e.g. porcelain restorations and multilayered composites.

42.3 CAD-CAM dentistry

There are a number of advantages to the use of digital restorations, notably time savings compared with traditional laboratory-fabricated restorations and fewer patient visits. Minor adjustments, generally, are easily made without the need for reglazing or heat treatment and/or recasting of metals. Since a partial coverage restoration, e.g. an inlay or onlay, can be fabricated instead of a full coverage crown, digital restorations can be more conservative of tooth structure and potentially of lower cost in many clinical situations. It is also possible to refine margins, contacts, etc. on the CAD pattern before the finished file is sent to the CAM unit for processing.

On the other hand, there is a learning curve with this technology so that “one-visit” dentistry may not always be possible, particularly if multiple postfabrication adjustments are necessary. Likewise, complex three- or four-surface restorations may require more computer design time and longer milling than simpler restorations.

Sometimes, digital restorations are approximations, not exact matches to the patient's teeth, and an accurate bite cannot always be guaranteed, which can compromise esthetics and cause an unbalanced occlusion. However, software updates and more comprehensive data banks are being developed almost daily. Finally, the technology for CAD-CAM restorative procedures does not always guarantee good marginal fit and the ceramics (and resins) used for digital restorations cannot be swaged like gold crowns.

42.3.1 Restoration luting

Digital restorations usually are luted with dual-cure adhesives although if sufficiently translucent, light-cured adhesives can be used. Light-cured GICs and RMGIs are not recommended for ACRs because hygroscopic expansion can cause restoration fracture.

It has been reported that the marginal gap with digital restorations is typically 60–150 μm , i.e. akin to those with conventional (cast) restorations. If the luting material can be leached out or worn away, then it is possible for ditching to occur around the restoration.

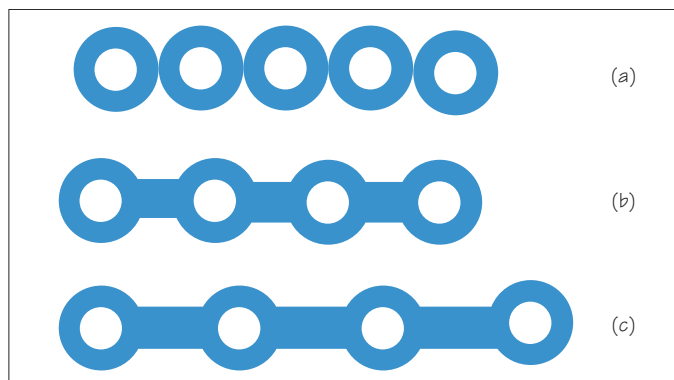


Figure 43.1 Schematic diagram of orthodontic elastomeric chains (A, closed loop; B, short open loop; C, long open loop).

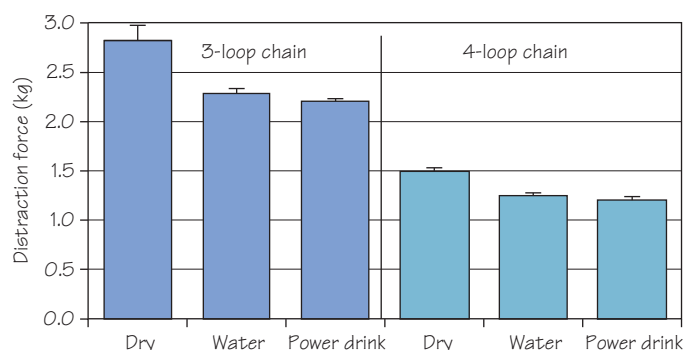


Figure 43.2 Force required to distract three- and four-loop segments of elastomeric chain in the dry (as-received) condition and after immersion in water and a citrate-containing power drink.

Table 43.1 Appliances for jaw orthopedics

Device	Application
Headgear	Exerts a specified amount of pressure on the maxilla to ensure proper eruption of upper teeth and guide maxillary growth direction.
Bionator	Assists both jaws to grow in proportion to one another, usually keeping the mandible forward and promoting proper tooth eruption.
Herbst	Semipermanently affixed to both sets of molars to prevent recessive mandibular movement and protruding maxillary teeth.
Frankel	Combines functional orthodontic therapy of the mandible while correcting tooth positioning in the maxilla.
Expansion devices	A palatal expansion device is applied to the maxillary teeth to help expand the maxillary width and reduce or eliminate crossbite .

Table 43.2 Characteristics of common orthodontic arch wires

Alloy	Composition	Hardening mechanism	Modulus	Formability	Solderability	Springback
Stainless steel	Fe-18%Cr-8%Ni-0.2%C	Cold work and interstitial carbon	High	Excellent	Difficult to solder; joints corrode	Adequate
Elgiloy	40%Co-20%Cr-15%Ni-7%Mo-16%Fe	Cold work and solution hardening	High	Excellent	Difficult to solder	Adequate
Nitinol	52%Ni-45%Ti-3%Co	Solid state solution and cold work	Low	Limited	Cannot solder	Excellent
Beta-titanium	Ti-1%Mo-6%Zr-4%Sn	Phase transformation	Intermediate	Excellent	Solderable	Excellent

Table 43.3 Yield strength, elastic modulus, and YS/E ratio for orthodontic arch wires

Alloy	Yield strength (GN/m ²)	Elastic modulus (GN/m ²)	YS/E ratio ($\times 10^{-2}$)
Stainless steel	0.3–1.9	160	1.1
Elgiloy			
AR	1.03	200	0.5
HT	1.77	200	0.9
Nitinol	1.66	33	5.0
Beta-titanium	1.17	65	1.8

AR, as-received; HT, heat-treated.

43.1 Intraoral appliances

Orthodontic appliances, used to correct **malocclusions** through tooth and jaw movement, fall into two broad classifications. **Fixed appliances** use brackets bonded to teeth whereas removable appliances use active elements contained in an acrylic resin base. **Removable appliances** can be **active** or **passive**, the former moving teeth while the latter (retainers and space maintainers) maintain tooth positioning.

43.2 Extraoral appliances and jaw orthopedics

Extraoral and **intraoral** appliances are used for **jaw orthopedics** (Table 43.1). They actively guide growth and development of the jaws to make the lengths of the mandible and maxilla compatible.

43.3 Removable orthodontic appliances

Removable appliances, used for limited maxillary orthodontic correction, comprise active elements (wires, springs, screws) that exert orthodontic forces and retentive elements (clasps) for retaining the appliance in the mouth; an acrylic base holds the two sets of elements together. **Space maintainers** assist normal eruption of teeth following premature tooth loss, injury, or other problems. **Retainers** prevent relapse of teeth or jaws to their original positions after tooth movement is complete.

These laboratory-fabricated appliances are inexpensive, relatively simple to adjust, require little clinical time, and can be removed for cleaning, during contact sports, or if the appliance causes irritation. Their disadvantages include bulkiness and the need for patient co-operation and habituation. Treatment is slower than for fixed therapy and removable appliances are unsuitable for mandibular treatments. Only tipping movements are possible, and removable appliances cannot perform tooth rotation.

A new orthodontics approach uses a series of clear plastic **aligners** fabricated by **CAD-CAM technology** to progressively move teeth without requiring brackets. As tooth movement progresses, the aligners are replaced at regular stages throughout therapy. Aligners are used primarily for simple malocclusions due to limitations in what can be achieved.

43.4 Fixed orthodontic appliances

Fixed appliances dominate orthodontic treatment. **Brackets** attached to the teeth are made of metal, ceramic, or plastic (clear or colored polycarbonate), comprising a base, a **slot** (and sometimes a tube) to carry the **arch wire**, and **ligation wings**. The arch wires that activate tooth movement are ligated into the bracket slots with fine stainless steel wire or rubber loops hooked over the wings.

Traditionally, brackets were spot-welded to stainless steel bands encircling the teeth but modern therapy uses brackets with **mesh-backed** or **perforated bases** that are bonded directly to the teeth. Brackets customarily are bonded to the buccal surface although lingual placement is sometimes used.

Therapeutic approaches utilizing fixed appliances include edgewise, Begg, straight wire, and Ricketts techniques, all approaches utilizing activate elements for tooth movement. Active elements may be arch wires, coil springs, and complex springs (e.g., the Jasper jumper and Forsus springs) as well as elastomeric chains. Bracket design, particularly the **slot configuration**, varies with the therapeutic technique, with some orthodontists designing their own brackets to achieve particular treatment goals. In many of these modalities, tooth movement is determined by the applied force, bracket positioning on the tooth, and/or the slot configuration.

The arch wires used in fixed-appliance therapy may be circular, rectangular, or square in cross-section. Wire selection, both the alloy and the wire shape, is based upon the therapeutic technique and the required activation force. Characteristics of the commonly used arch wires are indicated in Table 43.2.

Activation (i.e., tooth movement) is achieved by relaxation of stresses incorporated into the wire through bends and loops. Alternatively, bracket positioning may be varied so that the spring element moves the bracket and tooth into the required position. **Wire spring-back**, the maximum **elastic deflection** and the force used for tooth movement, is proportional to the ratio of yield strength to elastic modulus (**YS/E**; Table 43.3). Arch wire activation increases with the YS/E ratio and is the primary determinant in fixed appliance therapy. The optimum applied loads for tooth movement appear to be in the range of 150 grams.

Polyurethane-based **elastomeric chains** hooked over wings on the brackets are increasingly used for tooth movement because of convenience, effectiveness, cost, and patient acceptability. These elastomeric materials are available in closed, short open, and long open chain configurations (Figure 43.1). The **chain morphology**, the number of loops, and the chain color affect the **distractibility** of the chains, that is, the force that the chain will exert (Figure 43.2). Chain configuration selection is based upon the required gap closure and the activating force to initiate and sustain tooth movement.

Although the initial force applied by chains is high, they lose up to 50% of their tensile strength within 24–48 hours following placement. **Prestretching** of the chains before use produces more predictable load-extension behavior. Saliva, mouth washes, and certain soft drinks can plasticize the elastomer, reducing the force required to extend the chains and, conversely, the force delivered by a certain chain extension (Figure 43.2). This results in less predictable elastic behavior, particularly at higher applied loads.

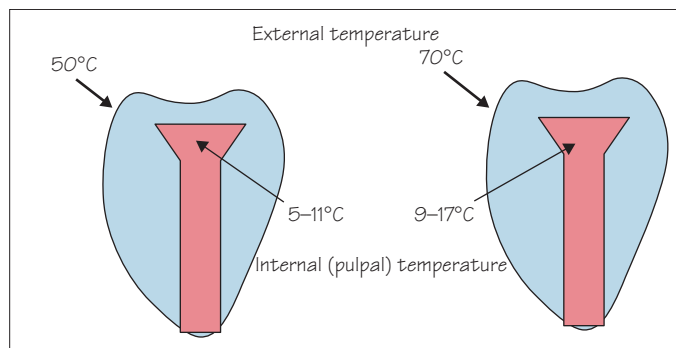


Figure 44.1 Effect of external temperature on internal (pulpal) temperature of teeth.

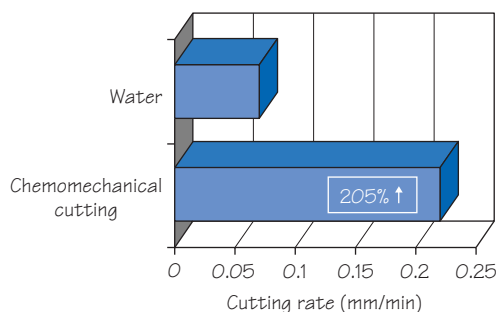


Figure 44.3 Chemomechanically enhanced dental cutting.

Box 44.1 Effects of cutting, grinding, and polishing on surfaces

Surface material is removed.
Subsurface layers are affected.
Microcracks are generated.
There is a nonuniform stress distribution.
An anisotropic surface layer is created.

Table 44.2 Abrasive particle and grit sizes

Classification	Particle size (grit)	Particle size (μm)
Extra coarse	60–100	>125
Coarse	120–140	105–125
Medium	160–200	70–90
Fine	220–280	50–65
Superfine	300–600	30–45
Microfine	1000–2000	10–20
Diamond paste	>14,000	<1

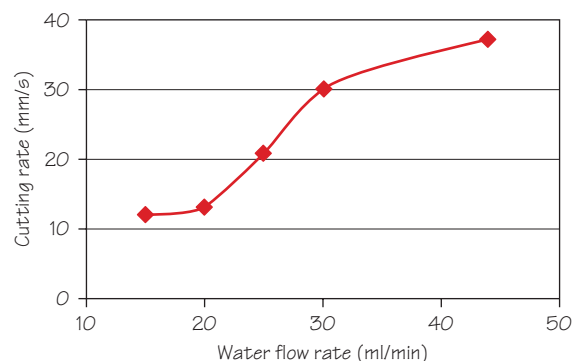


Figure 44.2 Effect of coolant flow rate on dental cutting rates.

Table 44.1 Common abrasives

Abrasive	Composition	Applications
Alumina	Aluminum oxide	Finishing of metals, resins, and ceramics
Arkansas stone	Microcrystalline quartz	Fine grinding of enamel and metals
Chalk	Mineral form of calcite	Polishing of enamel, amalgam, and resins
Corundum	Alumina	Grinding of metals
Diamond	Natural	Finishing and polishing of ceramics and resins
	Synthetic	Cutting, finishing, and polishing of tooth structure, ceramics, metals, and resins
Emery	Fine-grain alumina	Finishing of metals and resins
Garnet	Silicate-based mineral	Grinding of metals and resins
Pumice	Volcanic material	Polishing of enamel, amalgam, and resins
Quartz	Silica	Finishing of metals
Rouge	Iron oxide	Polishing noble metals
Sand	Silica minerals	Air blasting to clean castings
Silicon carbide	Synthetic material	Cutting of metals, ceramics, and resins
Tin oxide	Metal oxide	Polishing teeth and metals
Tripoli	Sedimentary rock	Polishing of metals and resins
Zircon	Zirconium silicate	Prophylaxis paste

Table 44.3 Cutting action of carbide and diamond burs and bur selection for different substrates

	Carbide burs	Diamond burs
Cutting action	Surface plowing Plastic deformation in flow zone Strain occurs Dislocations accumulate Cracks nucleate, grow, and propagate Interacting cracks form chips	Dislocation movement Outward flow of material Plastic strain limited Stress generation in near-surface region Cracks form and coalesce Interacting cracks form chips
Substrate surface hardness	Faster cutting with lower hardness	Faster cutting with greater hardness
Substrate	Better for ductile materials	Better for brittle materials
Enamel	Slower cutting	Faster cutting
Dentin	Faster cutting	Slower cutting
Sectioning base metals	Faster cutting	Slower cutting
Sectioning palladium alloys	Slower cutting	Faster cutting
Sectioning gold alloys	Slower cutting	Faster cutting
Sectioning titanium alloys	Faster cutting	Slower cutting

Abrasives and rotary cutting instruments (burs) are used for caries removal, cavity preparation, and margination of teeth as well as finishing and polishing of restorations.

44.1 Material removal

In engineering, drilling and lathe cutting involves chip formation but dental “cutting” is actually a **grinding process** with material being removed by **abrasion**. During material removal, bond rupture and molecular rearrangements change the substrate surface (Box 44.1). Polished metal surfaces become covered by the amorphous, less reactive **Beilby layer**, which is akin to the **smear layer** that forms on cut dentin.

44.1.1 Abrasives

Abrasives (Table 44.1) are classified by particle size or grit size (Table 44.2). These two characteristics are inversely related, larger particles having smaller grit sizes.

The abrasive particle size determines the **depth of scratch** in the surface, the effects of which extend significantly beyond the scratch itself; this affected region is termed **cold worked**. During surface smoothing, areas of unequal stress distribution are created and affect susceptibility to wear, fracture, and corrosion. Typically, finishing of a restoration or tooth material with a rapidly rotating bur generates defects (e.g., **microcracks**) to a depth of $\leq 50\mu\text{m}$, this region being less resistant to abrasion, wear, and corrosion.

44.1.2 Dental burs

Rotary cutting instruments are used in handpieces and are available in a variety of shapes. They comprise tungsten carbide (WC) and diamond burs (“diamonds”) with different grit sizes. Ultrahigh-speed air turbine handpieces have high speeds (300,000–400,000rpm) but poor torque, whereas low-speed handpieces are slower (30–100,000rpm) but can be used under high-torque conditions. The moderate-speed, heavier, and more bulky electric handpieces have higher torques than air turbines.

1 Tungsten carbide (WC) burs: WC burs generally are used for operative intracoronary procedures whereas diamonds are used for extracoronary procedures. Preparation refinement, margination, trimming, and polishing of restorations, etc. may be performed with 8-, 12-, 16-, 20-, or 30-fluted WC finishing burs.

2 Diamond burs: Diamond burs have a central metal shaft or spline to which the diamond abrasive particles are attached by electroplated metal (usually nickel) or the diamond particles are dispersed in a metal (commonly bronze) matrix that is sintered onto the spline. Sintered burs generally have smaller particle size diamonds and are slower cutting than plated burs but leave a finer surface finish. Finishing burs have medium- or fine-grit (8–40 μm) diamonds.

3 Carbides versus diamonds: Carbides and diamonds differ in cutting action (Table 44.3). Carbide burs are cheaper but less durable than diamond burs; disposable diamonds are markedly cheaper than conventional (multiuse) instruments and rival WC burs in cost.

4 Cutting efficiency: Bur cutting efficiency decreases with use and there is no difference in cutting efficiency between medium- and coarse-grit burs or between **disposable and conventional** diamonds. Initial WC cutting efficiency is greater than for diamonds but decreases more rapidly with use. There appears to be little difference in finish or properties of restorations when finished with WC or diamond burs. Diamond tools are used in the shaping of CAD-CAM ceramic restorations and for their adjustment, although carbides can be used for adjustments.

44.2 Coolant and lubricant effects

Dental cutting should be performed under water spray since elevated temperatures at the tooth surface can cause severe and sometimes irreversible pulpal damage due to internal temperature rise (Figure 44.1). Cutting rates are faster with greater coolant flow rates (Figure 44.2).

Recent studies indicate that faster cutting with better surface finish, less restoration leakage, and longer bur life can be achieved through **chemomechanical effects** (Figure 44.3). The latter are possible through low additions of monohydric and trihydric alcohols to the handpiece coolant.

Polishing must be done at slow speed and the temperature kept low when trimming or polishing polymeric materials to avoid thermal distortion of the resin matrix. The T_g of PMMA is ca. 105°C and heating effects from polishing will cause distortion, e.g. of dentures, as the temperature approaches the T_g .

44.3 Finishing and polishing regimens

Chairside and extraoral laboratory trimming, adjusting, and polishing procedures should always be performed in sequence from the coarsest grit to the finest.

Finishing discs enable more precise operations than WC or diamond burs and often with greater safety. Most discs and strips have alumina abrasives although devices with silicon carbide and diamond abrasives also are available. They are supplied in color-coded grit sizes from coarse to fine/superfine.

Finishing WC burs and diamonds appear to be equally popular, although diamonds are preferred for contouring, adjusting, and smoothing of porcelain. WC finishing burs are typically used dry whereas diamonds are used under water cooling.

Rubber polishing instruments with embedded diamond particles are used extensively whereas polishing paste use has diminished. These tools, predominantly used at low speeds, appear to be more effective when used dry than wet although excessive heat can be generated if not used with care.

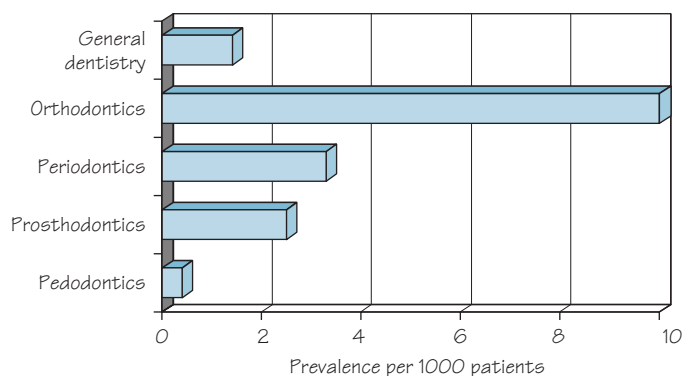


Figure 45.1 Prevalence of adverse reactions in dental specialties.

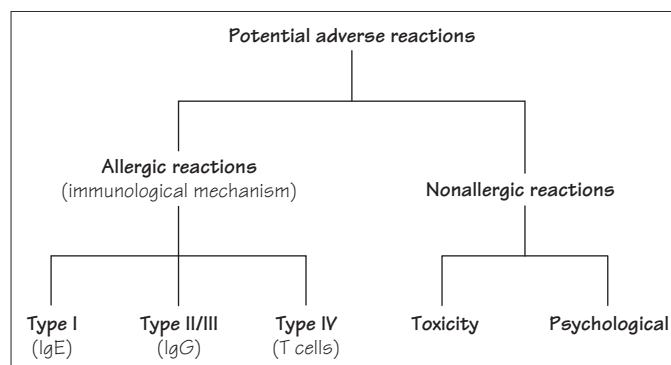


Figure 45.2 Potential adverse reactions.

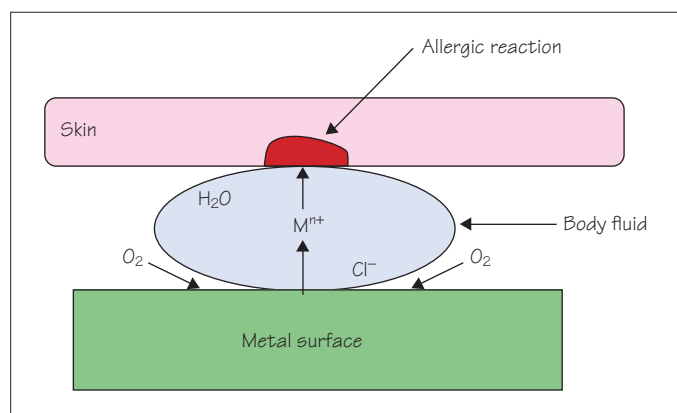


Figure 45.3 Corrosion of metal by reaction with sweat or saliva.

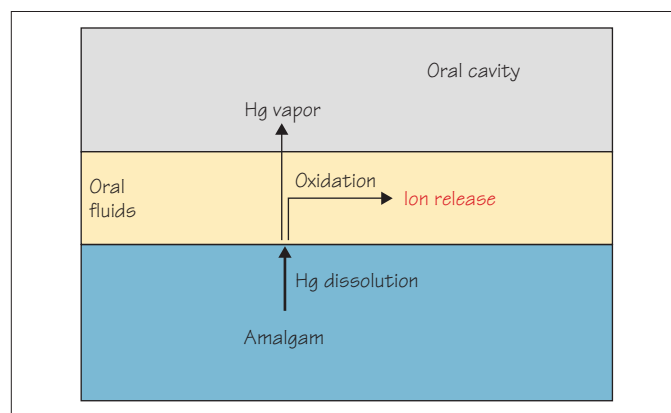


Figure 45.4 Mechanism of mercury release from amalgam restorations.

Table 45.1 Carcinogenic and toxic substances used in dentistry

Carcinogens	Toxic substances
Formaldehyde	Methyl methacrylate
Cadmium	Formaldehyde
Beryllium	Mercury vapor
Metals:	
	Arsenic
	Barium
	Cadmium
	Chromium
	Lead
	Mercury
	Selenium
	Silver

Table 45.2 Principal allergic reactions

Allergy	Reaction type	Mechanism
Type I	Anaphylactic reactions—antibody (IgE) mediated	Immunoglobulin antibodies (IgE) bind to receptors on mast cells. Pharmacologically active compounds may be released. Clinical effect may be respiratory system obstruction and cardiovascular collapse.
Type II	Cytolytic or cytotoxic	Immunoglobulins (IgM or IgG) bind to antigens on surface of cells and activate complement. Activation may result in cytolysis, phagocytosis, and chemotactic reactions.
Type III	Immune-complex	Occur when complexes made of IgM and IgG antibodies accumulate in blood vessels or tissue and activate the complement system
Type IV	Delayed-type hypersensitivity (T-cell mediated)	Immune response is mediated by T cells, usually CD4+. Cytokines are released, leading to macrophage activation and resulting in local damage.

Ig, immunoglobulin; IgE, immunoglobulin E; IgG, immunoglobulin G; IgM, immunoglobulin M.

Clinical use of dental biomaterials, whether natural or artificial, is determined by function, properties, and any associated risk. There is probably no material without the potential for causing biocompatibility problems.

Adverse reactions occur only when material is released into the biosystem; for example, salivary contact with biomaterials can cause

corrosion and release metal ions whereas low molecular weight compounds may be released through leaching and hydrolysis. The potential effects of luting agent swelling through water sorption, i.e. hygroscopic expansion, on ceramic restorations were mentioned in Chapter 29 and Chapter 42.

Table 45.3 Type I and Type IV allergic reactions

	Type I	Type IV
Type of reaction	IgE-mediated anaphylactic	T-cell mediated, delayed-type hypersensitivity
Characteristic	Immediate hypersensitivity	Delayed hypersensitivity
Initiator	Specific antigen contact	Antigen-sensitized T cells
Mechanism	Humoral, cutaneous, or systemic reactions	Contact reaction to chemicals or acquired immune reaction
Response time	Minutes to hours	Minutes to hours
Duration	Diminishes within hours	Can persist for weeks
Examples	Allergic asthma and rhinitis; food and drug reactions; reaction to insect bites	Contact dermatitis; allograft rejection

IgE, immunoglobulin E.

The prevalence of adverse reactions is **1.4 per 1000** patients or in 0.15% of patient-clinician encounters, although it is higher for certain specialties (Figure 45.1). Overall, the prevalence of adverse dental reactions is the same as that for adverse pharmaceutical reactions.

Adverse reactions fall into two broad categories (Figure 45.2), the most common being hypersensitivity (allergic) reactions.

45.1 Toxicity and carcinogenicity

Many toxic and carcinogenic “dental” compounds exist (Table 45.1), but they rarely elicit cancer and toxic reactions.

45.2 Hypersensitivity and allergic reactions

At least 10% of the population suffers from one or more allergies but only 1 in 3 shows a positive reaction to patch tests. Most people experience no inconvenience from their allergies.

45.2.1 Hypersensitivity

Hypersensitivity is an increased reaction against stimuli, initial symptoms starting in skin, mucosa of the eyes, respiratory tract, or the GI tract. It results from specific reactions between antigens and humoral antibodies or sensitized lymphocytes (immunologically active cells) and is initiated by high local concentrations of an irritant or a small amount of an immune system activator.

45.2.2 Allergic reactions

Allergic reactions are induced by **allergens**, the hypersensitivity reaction typically being histamine release, swelling of the mucous membrane, sneezing, or itching. An allergy therefore is a strong immunological reaction with a prerequisite of previous contact with lymphocytes. Four principal types of allergic reactions are known (Table 45.2).

Type I and Type IV reactions (Table 45.3 and Table 45.4) are the most common in dentistry, particularly **contact allergy**. The latter can occur with clinicians and patients, but allergic contact stomatitis is uncommon.

Most allergens inducing **allergic contact dermatitis** are low molecular weight compounds (**haptens**) that conjugate with proteins. The resulting complex (**antigen**) induces sensitization of immunocompetent cells. The supramolecular chemistry of the hapten–protein

Table 45.4 Common allergens in dentistry

Clinicians	Patients
Latex gloves	Nickel
Methacrylates	Latex
Disinfectants	Impression materials
Anesthetics	Rubber compounds
Metals	Mercuric chloride
Colophony	Cobalt salts
Eugenol	Thimerasol
Impression materials:	Benzoyl peroxide
Polyethers	Fragrances
Alginates	
Gluteraldehyde	

Table 45.5 Prevalence of metal allergies

Metal	Females (%)	Males (%)
Nickel*	39	3
Chromium	1.5	2
Cobalt	1	1
Mercury	2	2

*Most persons allergic to nickel are also allergic to palladium; pierced skin increases the prevalence of nickel allergy.

complex determines antigen binding to receptors on T lymphocytes and its ability to cross the dermal barrier.

There is little evidence of adverse reactions to resin-based restorative materials although there are infrequent cases of reactions to PMMA dentures.

45.2.3 Metal allergies

Metals enter the body by various routes and metal allergies due to corrosion from dermal contact (Figure 45.3) with sweat or saliva are relatively common. The prevalence of common metal allergies is shown in Table 45.5.

The high prevalence of adverse (allergic) reactions in orthodontic patients, 85% of cases being associated with extraoral anchorage, is ascribable to the stainless steel in these devices. Likewise, 27% of adverse reactions in prosthodontic patients are related to base metal alloys.

45.2.4 Reaction to mercury

Mercury sensitivity (by patch test) is found in 2% of the population. Although mercury can be released from amalgam restorations (Figure 45.4), there have been only 50 reported cases of allergic reactions in dental patients since 1906. It appears that insufficient mercury is released from amalgam to elicit an immunological reaction.

Pulpal reactions observed with amalgam restorations placed on freshly cut dentin are ascribable to condensation pressure; similar reactions have been found with gold foil. Overall, dental amalgam appears to be a safe restorative material with a number of clinical advantages.

45.3 Conclusion

Adverse reactions to dental materials are very infrequent; there is a greater prevalence of allergic/hypersensitivity reactions due to perfumes, cosmetics, cleansers, and detergents than from dental biomaterials. Nevertheless, caution should be exercised when using nickel-containing alloys for patients with a history of metal-related allergies.



Figure 46.1 Enamel dissolution caused by excessive consumption of citrus soft drinks. (Courtesy of Dr. Howard Strassler.)



Figure 46.2 Enamel dissolution caused by consumption of a sports drink. (Courtesy of Dr. Howard Strassler.)

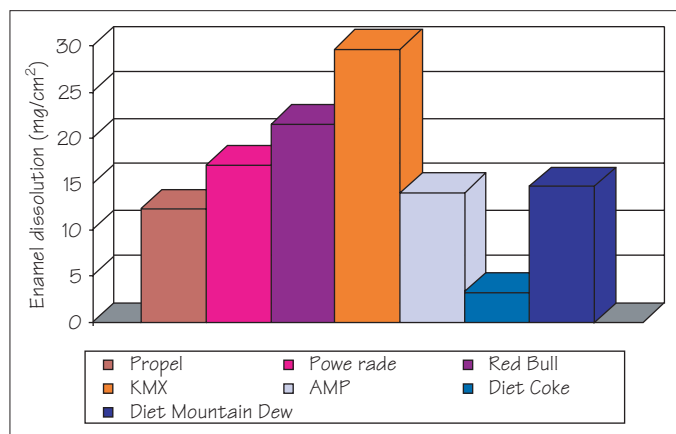


Figure 46.3 Enamel dissolution in soft drinks (14 days).

Dental erosion is irreversible hard tissue loss that does not involve bacterial, mechanical, or traumatic factors (Figure 46.1 and Figure 46.2). Erosion is associated with intrinsic and extrinsic acids.

Intrinsic acids are gastric acids present in the mouth due to induced vomiting, e.g. in bulimics and alcoholics, and reflux acids caused by GERD. Erosion of lingual surfaces is often diagnostic of bulimia and GERD.

Extrinsic acids derive from an acidic diet, commonly wines and low-pH beverages with a high organic acid content. Low-pH foods, e.g. pickles, fresh fruits, and yogurt, can contribute to dental erosion.

46.1 Soft drink consumption

Soft drink consumption is increasing by 2–3% per year. It has risen sevenfold over the past 50+ years in the United States and by 56% in the United Kingdom in the past 10 years, with similar trends likely in other countries.

Sugary beverage intake is higher in adolescents and children than adults, diet drinks possibly being more popular with the latter. The consumption of noncola (citrus-flavored) beverages is growing rapidly.

Dental erosion rates are the same for regular and diet drinks; fruit-flavored drinks and unsweetened juices have the same erosivity as carbonated beverages.

Erosion is higher in children/adolescents from higher socioeconomic groups; lower socioeconomic groups have more plaque and caries but limited erosion. Whereas plaque deposits predispose teeth to caries and periodontal problems, they may provide some protection against accretion of organic acids and subsequent erosion.

46.2 Beverage acidity

The protective pH level of saliva is 5.5; enamel is at risk of demineralization if the oral pH falls below this level. After acid exposure, the oral cavity takes approximately 30 minutes to revert to the safe pH range; prolonged or repeated exposure to a lowered pH accelerates enamel demineralization.

The **titratable acidity (TA)** of a beverage measures the total acid level and the actual hydrogen ion availability for interaction with teeth. TA is predictive of the erosive potential of wines and beverages; carbonated beverages have high TA values because acids are added as taste enhancers and to offset sugar sweetness. Many acidic additives, notably citrus flavorants, are strong buffers and maintain low oral pH levels even when markedly diluted, enhancing erosivity (Figure 46.3).

Temperature is also important in that acids are less erosive cold than warm; allowing acidic drinks to remain in the mouth and warm up increases their erosivity.

46.3 Acids in beverages

Common organic acid flavorants include citric, malic, lactic, and tartaric acids, most of which occur naturally in foodstuffs. Citric acid and

citrates are found in orange, lemon, and other fruit juices. Malic acid is present in apples and fruit juices. Tartaric acid and various tartrates are found in wine and grapes. Lactic acid is present in milk, cream, and fermentable foodstuffs such as yogurts, sauerkraut, and cream sodas.

Phosphoric acid is common in colas because it is a strong but relatively low-cost acidulant. Despite being about three times more erosive than organic acids, phosphoric acid lacks buffering capacity so that beverage residues in the mouth are rapidly neutralized and, overall, there is less erosive attack than with less aggressive but highly buffering organic acids.

Enamel erosion from wine drinking is known, particularly with wine tasters and consumers who like to “swish” it around the mouth regularly. Wine tasters sample 5–50 wines a day and hold the wine in their mouths for 15–60 seconds, sufficient time to cause significant enamel erosion.

46.4 Factors in erosion

Chemical, biological, and behavioral factors affect susceptibility to erosion. Chemical factors include beverage pKa values (which determine acidic strength), chelating properties, and adhesion of additives to the enamel surface. Biological factors include the flow rate, buffering capacity, and composition of saliva, pellicle formation, and the dental tissues. In particular, the calcium, phosphate, and fluoride contents of enamel influence susceptibility to erosion. Deciduous teeth, inherently more porous than permanent teeth, have greater susceptibility to acid attack.

Behavioral factors include eating and drinking habits, lifestyle, and excessive or prolonged consumption of acidic beverages. However, protective effects are found with foods containing calcium, phosphate, and fluoride; even yogurt consumption is beneficial.

46.5 Other causes of erosion

Swimmers, subject to repeated ingestion of chlorinated pool water, also show evidence of enamel erosion. Environmental exposure to acidic gases in the work environment, e.g. in electroplating and acid pickling plants, can cause dental erosion because the oral pH is lowered with repeated exposure to such gases.

Many “street” drugs, notably methamphetamines, cocaine, and ecstasy, directly or indirectly cause severe enamel erosion. Common clinical signs of drug abuse include erosion and gingival recession (e.g., with cocaine). Severely worn and decayed maxillary anterior as well as posterior occlusal wear are common with methamphetamine users. Street drugs also increase the body’s need for adenosine triphosphate (ATP) and users tend to consume carbohydrates and carbonated drinks to satisfy increased energy demands.

Glossary

- >: Mathematical symbol for “greater than” (e.g., $a > b$ denotes that a is greater than b)
- <: Mathematical symbol for “less than” (e.g., $a < b$ denotes that a is less than b)
- $\leq$: Mathematical symbol for “equal to or less than” (e.g., $a \leq b$ denotes that b is equal to or less than a)
- $\geq$: Mathematical symbol for “equal to or greater than” (e.g., $a \geq b$ denotes that a is equal to or greater than b)
- $\sim$: Mathematical symbol for “about”
- 4-META**: 4-Methacryloxyethylenetrimellitic anhydride (dentin-bonding agent)
- Abrasion**: Wear between two moving surfaces in contact
- ACR**: All-ceramic restoration
- Acrylic resin**: Resin formed by polymerization of methyl methacrylate
- Adhesion**: Bonding between unlike molecules such as an adhesive to a surface
- Adhesion quotient**: Amount by which the substrate surface energy (γ_{SA}) exceeds the adhesive surface tension (γ_{SL})
- All-ceramic restoration**: Material containing up to 99 vol.% of a reinforcing crystalline phase
- Allergic reaction**: Hypersensitivity induced by allergens
- Aluminous porcelain**: Porcelain strengthened by addition of powdered alumina
- Amalgam alloy**: Silver–tin alloy particles that react with mercury to form dental amalgam
- Anaphylactic reaction**: Type I immunologic reaction
- Anelastic recovery**: Recovery of inelastic strain after release of applied loading
- Anodyne**: Pain-relieving
- APF gel**: 1.23% acidulated phosphate fluoride gel, a topical fluoridation agent.
- Beilby layer**: Featureless, less reactive layer formed on surfaces by polishing
- bis-Acryl (bis-acrylic)**: Hybrid acrylic resin composite often used as a temporization material
- bis-GMA**: Bowen’s resin
- Blue light**: 400–500 nm wavelength light used to cure light-curing resins
- BPBG**: Butyl phthalyl butyl glycolate (a plasticizer)
- BPDM**: Biphenyl dimethacrylate (dentin-bonding agent)
- Burnishing index**: Ability to be worked in the mouth or burnished (ratio of percentage elongation to yield strength)
- ca.:** Circa—an abbreviation for “about” or “approximately”
- CAD-CAM**: Computer-aided design–computer-aided manufacturing
- Cameo surface**: Polished surface of a denture that faces into the oral cavity.
- Camphoroquinone (CQ)**: Photoinitiator for resins
- Cavity liner**: Suspension of calcium hydroxide [$\text{Ca}(\text{OH})_2$] in solvent
- Cavity varnish**: Barrier coating for dentin against fluid penetration
- CD**: Complete denture
- Ceramic**: Porcelain
- CEREC**: CERamic REConstruction, a commercial CAD-CAM system for fabricating ceramic restorations
- Ce-TZP**: Ceria-stabilized tetragonal zirconia polycrystals
- CFU**: Colony-forming unit—an estimate of viable bacterial or fungal numbers
- Chemomechanical effect**: Accelerated cutting via reagents added to the handpiece coolant water
- China clay**: Kaolin ($\text{Al}_2\text{O}_3 \cdot 2\text{SiO}_2 \cdot 2\text{H}_2\text{O}$)
- CIE**: Commission International de l’Eclairage; a color-measuring system
- CMC**: Carboxymethyl cellulose (thickening agent)
- Cohesion**: Chemical bonding between like molecules
- Compomer**: Polyacid-modified composite containing fluoride-releasing silicate glass
- Composite**: Solid material with a resin matrix and filler particles
- Compressive strength**: Load to failure divided by cross-sectional area under compressive loading
- Condensation**: Removal of water from the porcelain powder–water slurry
- Conditioning**: Creation of surface topography
- Conventional composite**: Composite containing larger filler particles
- Corrosion**: Loss of metallic characteristics of metals due to an electrochemical reaction
- Coupling agent**: Silane used to promote adhesion between resin and silica filler
- C.P. (c.p.)**: Abbreviation for “commercial purity” or “commercially pure”
- Critical surface energy (CSE)**: Value of surface tension between solid and liquid when liquid just spreads over the surface
- CSE**: Critical surface energy
- CTE**: Coefficient of thermal expansion; change in length per unit length for 1°C temperature change
- DBA**: Dentin-bonding agent
- DBP**: Dibutyl phthalate (a common plasticizer)
- Delayed-type hypersensitivity**: Type IV immunologic reaction
- Dental cement**: Luting agent used to “cement” a restoration to tooth structure
- Dental cutting**: Grinding process with material being removed by abrasion
- Dental erosion**: Loss of dental hard tissue by a process not involving bacteria (i.e., chemical dissolution)
- Dental glass**: Dental porcelain containing little or no kaolin
- Devitrification**: Crystallization of the glass matrix of porcelain
- Dibutyl phthalate**: Plasticizer (softener) for resins
- Dispersion-modified alloy**: Amalgam alloy comprising admixture of silver–tin and silver–copper particles
- DMFT**: Decayed, missing, and filled teeth score
- Ductility**: Percentage elongation ($\Delta L/L \times 100\%$)
- E4D**: Commercial CAD-CAM system for restoration fabrication
- EBA**: Orthoethoxybenzoic acid (a strengthener for ZOE cement)

- EDTA:** Ethylene diamine tetraacetic acid (dentin conditioner)
- Elastic limit:** Maximum stress that can be applied without permanent deformation
- Elastic modulus:** Ratio of stress to strain (also known as modulus of elasticity or Young's modulus)
- Enamel erosion:** Loss of dental enamel by a nonbacterial process (i.e., chemical attack)
- Feldspar:** Potassium aluminum silicate (orthoclase) and sodium aluminum silicate (albite)
- Feldspathic porcelain:** Porcelain formed using feldspar as the flux
- Firing:** Sintering to form glass bridges that flow between unfused porcelain particles
- Fracture toughness:** Critical value at which fracture occurs when it is exceeded by the stress intensity
- Frit:** Powder formed by quenching and grinding of reaction product of feldspar with silica, kaolin, or glass
- Gamma phase (γ phase):** Ag_3Sn particles in amalgam alloy
- Gamma-1 phase (γ_1 phase):** Silver-rich reaction product (Ag_2Hg_3) of mercury and the gamma phase
- Gamma-2 phase (γ_2 phase):** Tin-rich reaction product (Sn_{7-8}Hg) of mercury and the gamma phase
- Gantrez acid:** Poly[vinylmethylether maleate]—used in denture adhesives
- GERD:** Gastroesophageal reflux disease, a common cause of intrinsic oral acids
- GIC:** Glass ionomer cement
- Giomer:** Polyacid-modified composite containing fluoride-releasing silicate glass
- Glass ionomer:** Cement and restorative material based on silicate glass and polyalkenoic acids
- Glass transition temperature (T_g):** Temperature at which a rigid resin transforms to one with rubbery characteristics
- Glassy matrix:** Porcelain comprising dispersed alumina and quartz
- Glaze:** Low-fusing, transparent glass
- GLUMA:** Mixture of gluteraldehyde and hydroxyethyl methacrylate (dentin-bonding agent)
- Glycol dimethacrylate:** Cross-linking additive in polymerizing resins
- GMPTS:** Gamma-methacryloxypropyltrimethoxysilane (silanating agent)
- GP:** Gutta-percha
- Gypsum:** Naturally occurring dihydrate of calcium sulfate
- Hapten:** Low molecular weight compound that conjugates with proteins
- Hardness:** Resistance to penetration
- HEMA:** Hydroxyethyl methacrylate (dentin-bonding agent)
- Hemihydrate:** $\text{CaSO}_4 \cdot \frac{1}{2}\text{H}_2\text{O}$ (plaster of Paris)
- High-fusing porcelain:** Used for jacket crowns or denture teeth, consists of previously unfired feldspar, kaolin, and quartz
- High-impact resin:** Resin modified with elastomers to increase impact resistance
- HIP:** Hot isostatic pressing—method of fabricating components using powder metallurgy
- Hot-pressed ceramic:** Ceramic sintered and shaped by external pressure at high temperature
- Hybrid composite:** Composite containing two types of filler particle
- Hydrophilic:** Water-loving
- Hydrophobic:** Water-hating (i.e., nonwetting)
- Hydroxyethylmethacrylate (HEMA):** Facilitates coexistence of resin and acid in aqueous solution and participant in polymerization
- Hypersensitivity:** Increased body reaction against stimuli
- Imbibition:** Sorption of fluids by a gel on standing
- Incongruent melting:** Nonuniform melting of a substance with incomplete decomposition into another substance
- Indentation hardness:** Resistance to penetration
- Intaglio surface:** Tissue-fitting surface of a denture
- Interdigitation:** Merging or intermingling of molecules at a bonding interface by diffusion
- Investment material:** Hardenable material containing refractory, binder, and modifiers
- Ion strengthening:** Strengthening of porcelain by immersion in ion exchange bath
- Ionomer:** Abbreviation for “glass ionomer”—restorative material based on reaction of silicate glass with a polyalkenoic acid
- Irreversible hydrocolloid:** Alginate impression material
- Kaolin:** China clay ($\text{Al}_2\text{O}_3 \cdot 2\text{SiO}_2 \cdot 2\text{H}_2\text{O}$)
- Keesom forces:** Attractive forces arising from inherent positive and negative poles in the adjoining molecules
- Light-cure resins:** Resins containing photoinitiator for light cure
- London forces:** Attractive forces arising from poles induced by random electron motions
- Lost wax process:** Traditional casting process in which a wax pattern is used to create a casting mold
- Low-fusing porcelain:** Porcelain fabricated with a prefused, fritted powder
- Luting agent:** Dental cement
- MDP:** 10-Methacryloyloxydecamethylene phosphate (dentin-bonding agent)
- Medium-fusing porcelain:** Porcelain fabricated with a reduced melting point frit material
- Mer:** Single molecule of a polymer
- Metal-ceramic restoration:** Restoration with cast metal coping onto which porcelain is fired (PFM restoration)
- Metamerism:** Matching apparent color of objects with different spectral power distributions
- MMA:** Methyl methacrylate
- Monomer:** Unpolymerized methylmethacrylate
- MP:** Melting point
- MTA:** Mineral trioxide aggregate; restorative and endodontic material chemically similar to Portland cement
- Munsell system:** Color-measuring system using a large number of color tabs
- MW:** Molecular weight
- Newtonian fluid:** Fluid for which flow is proportional to applied load
- NPG-GMA:** *N*-Phenylglycine–glycidylmethacrylate (dentin-bonding agent)
- NTG-GMA:** Reaction product of *N*(*p*-tolyl)glycine and glycidyl methacrylate (dentin-bonding agent)
- Obtundent:** Drug or medication that has a sedative effect, i.e. lessens or relieves pain
- Oligomer:** Converted monomer with a finite degree of polymerization (i.e., a limited number of monomer units, usually 2–4 mers)
- Osseointegration:** Close apposition and adherence of bone to dental implants
- Overtrituration:** Overmixing of alloy–mercury mixture
- PEMA:** Poly(ethylmethacrylate)
- Penetration coefficient:** Determinant of adhesive penetration into etched enamel

- PENTA:** Phosphate penta-acrylate (dentin-bonding agent)
- PFM:** Porcelain-fused-to-metal restoration; also known as metaloceramic or ceramic–metal restoration
- P/L:** Powder-to-liquid ratio
- PMMA:** Poly(methyl methacrylate)—or acrylic resin
- PMDM:** Addition product of pyromellitic dianhydride and 2-hydroxyethyl methacrylate (dentin-bonding agent)
- Poisson's ratio:** Ratio of lateral to axial strain under tensile loading
- Polyalkenoic acid:** Polyacrylic, icatonic, and/or tartaric acid present in glass ionomer cement
- Polymer:** Solid (nonmetallic) high molecular weight compound composed of small repeating units
- Polymerization:** Reaction between free radicals and reactive centers in a monomer
- Poly(methyl methacrylate):** Acrylic resin
- Polyphosphazenes:** Wide range of hybrid inorganic-organic polymers with backbone of alternating P and N atoms with two organic or organometallic groups attached to each phosphorus atom
- Porcelain:** Material comprising crystalline minerals (silica, alumina) dispersed in a glass matrix
- Primer:** Dentin-bonding agent
- Proportional limit:** Maximum stress that a material can sustain without deviating from linear stress–strain proportionality
- Pyroplastic flow:** Slumping of porcelain when heated
- Rapid-cure resin:** Resin containing both chemical and heat-activated initiators
- Resilience:** Resistance to permanent deformation; energy required for deformation to the proportional limit
- Resinomer:** Polyacid-modified composite containing fluoride-releasing silicate glass
- Reversible hydrocolloid:** Agar–agar impression material
- Rheology:** Flow properties of liquids
- RMGI:** Resin-modified (hybrid) glass ionomer
- RPD:** Removable partial denture
- RT:** Room temperature
- Silicate glass:** Calcium fluoroaluminosilicate ($\text{SiO}_2\text{--Al}_2\text{O}_3\text{--CaF}_2\text{--Na}_3\text{AlF}_6\text{--AlPO}_4$) glass powder
- Sintering:** Coalescence (fusing without complete melting) of porcelain powder particles; firing of porcelain mass
- Slip-cast ceramic:** Condensed aqueous porcelain slip on a refractory die that is fired and infiltrated with molten glass
- Smear layer:** Featureless, poorly adherent layer of organic debris forming on cut dentin
- Specific heat:** Quantity of heat needed to raise the temperature of 1 g of substance by 1°C
- Strain:** Ratio of deformation to original length, $\Delta L/L$ —measures deformation at failure
- Stress:** Force per unit cross-sectional area
- Stress riser:** Small flaw or structural discontinuity at which applied stress is amplified and from which cracks propagate
- Syneresis:** Loss of moisture and components from a gel on standing
- TEGDMA:** Triethylene glycol dimethacrylate (monomer diluent)
- Tensile strength:** Stress under tension that causes failure
- Thermal coefficient of expansion:** Unit change in length for 1°C rise in temperature
- Thermal conductivity:** Rate of heat conduction through unit cube of material for temperature difference of 1°C across the cube
- Thermal diffusivity:** Thermal conductivity divided by specific heat multiplied by the density
- Thixotropy:** Property of fluid or semirigid gel that resists flow when static but allows flow upon agitation
- Titrateable acidity (TA):** Total acid level and actual hydrogen ion availability in a fluid
- Total etch technique:** Demineralization treatment of the dentin surface and subsurface
- Toughness:** Resistance to fracture (i.e., energy required to cause fracture)
- Transverse strength:** Strength of a material in flexion
- Tribochemical bonding:** Bonding pretreatment for ceramics—an alumina–silica layer applied to surface and then silanated
- Tristimulus values:** Three color components produced by eye's receptor cones by reduction of light wavelengths
- Trituration:** Reaction of mercury with silver–tin alloy particles to form dental amalgam
- UDMA:** Urethane dimethacrylate
- Ultimate strength:** Maximum stress sustained before failure
- Undertrituration:** Insufficient mixing of alloy–mercury mix
- van der Waals forces:** Attractive forces between two polar molecules based on charge differences
- Veneer:** Low-fusing, transparent glass
- Vitreous:** Quasi-supercooled liquid glass
- Vitreous phase:** Glass matrix or porcelain formed by a flux or binder such as feldspar
- Wetting:** Adhesive spread over the substrate
- W/P:** Water-to-powder ratio
- Yield strength:** Stress at which there is a specified deviation from proportionality of stress to strain
- Y-TZP:** Yttria-stabilized tetragonal zirconia polycrystals
- Zinc phosphate:** Traditional dental cement based on zinc oxide–phosphoric acid
- Zinc polyacrylate (polycarboxylate):** Cement based on zinc oxide and aqueous polyacrylic acid solution
- ZNC:** Zinc polyacrylate (polycarboxylate) cement
- ZNP:** Zinc phosphate cement
- ZOE:** Zinc oxide–eugenol cement

Index

- 4-META 63, 71
- Abrasion 5, 93
- Abrasion resistance 19, 21, 73, 77, 83
- Abrasives 92–93
- Acid-base reaction 60, 61
- Acidic drinks 96, 97
- ACRs (all-ceramic restorations) 39, 87
- Acrylic (methacrylate) allergy 43, 47
- Acrylic resin *see* Poly(methyl methacrylate)
- Addition-cured silicone 21, 30, 31, 33, 59
- Adhesion 8, 9, 54–55, 58, 59, 63, 64, 65, 68, 76, 78, 79
- Adhesion quotient 9, 11, 69, 89
- Adhesion zone 9
- Adhesive agent 9, 10, 11, 65, 71
- Adhesive dentistry 9, 68–69
- Adhesive layer 55
- Adhesive resins 69, 77
- Adhesive spread 63, 69
- Adhesives 55
- Advanced ceramic systems 86–87
- Adverse effects 65, 94–95
- Agar-agar 31
- AH26 75
- Alginate 29–30, 31
- Aligners 91
- Allergens 95
- Allergic reactions 43, 94–95
- Alumina 81, 84, 85, 86, 87, 92
- Aluminous porcelain 85
- Amalgam *see* Dental amalgam
- Ameloblast 12, 13
- Amelogenesis 13
- Anelastic recovery 51
- Antimicrobial agent 77, 79
- Apatite *see* Hydroxyapatite
- APF gel 13, 69
- Arch wires 90, 91
- Auto-cure (self-cure, cold cure) resin 45, 47, 59, 63, 73, 79
- Bacteria 52, 53
- Barcol hardness 3
- Barrier effect 57, 71
- Base metal alloys 36–37, 39
- Bases 56–57
- Beilby layer 93
- Benzoyl peroxide 44, 45
- Beryllium 36, 37, 39, 94
- Beta-titanium 90
- Bierbaum hardness 3
- Bioadhesives 65
- Biocompatibility 36, 37, 38, 41, 44, 54, 55, 56, 71, 74
- Bis-Acryl materials 33, 77
- Bis-GMA (Bowen's resin) 57, 63, 73
- Bisque 82
- Bite registration 32, 33
- Biting forces 3
- Bond failure 8, 54, 69, 71
- Bond strength 62, 63, 71
- Bone 14–15, 41
- Bone formation 14–15
- Bone grafting 15
- BPBG 51
- BPDM 63
- Brånemark 41
- Brinell hardness 3
- Brittle materials 3, 5, 19, 83
- Burnishing index 3
- Burs *see* Dental burs
- Ca(OH)₂ 78
- CAD-CAM technology 21, 33, 87, 88–89, 91
- Calcium hydroxide, Ca(OH)₂ 56, 57
- Cameo surface 49
- Camphoroquinone (CQ) 45, 73
- Candida 53
- Carat 35
- Carboxymethyl cellulose (CMC) 65
- Carcinogenicity 95
- Cast chromium alloys 36, 37
- Castable glass ceramics 86–87
- Casting 24, 25, 35, 37
- Casting defects 35
- Casting porosity 35
- Cavit® 77
- Cavity liner 56–57
- Cavity varnish 56–57
- Cement removal 59
- Cements *see* Dental Cements
- Cementum 13
- Ceramic etching 55
- Ceramics 5, 39, 86–87, 89, 90
- Ceramming 87
- CeraOne® 85
- Cermets (metal-reinforced GICs) 62
- Chemical adhesion 9, 61
- Chemomechanical effects 92, 93

- Chroma 7
 Chromium alloys 36, 37, 39, 43, 95
 CIE color system 7
 Citric acid 63, 70
 Cobalt-chromium alloys 36, 37, 39, 43, 90, 95
 Coefficient of thermal expansion (CTE) *see* Linear coefficient of thermal expansion
 Cohesion 8, 9, 55, 65, 68, 71
 Cohesive strength 54, 55
 Cold working 93
 Cold-cure resin *see* Auto-cure resin
 Collagen 12, 13, 71
 Colloids 19
 Colophony *see* Rosin
 Color 6, 7
 Complete dentures (CDs) 43, 44–45, 46–47, 65
 Compomer 63, 73
 Composite resin 63, 2–73, 77
 Compressive strength 5, 56, 60, 67, 72
 Condensation of porcelain 82, 83
 Condensation-cured silicone 30, 31
 Conditioning 69, 71
 Contact angle 11, 69
 Copal 23, 57
 Core build-up 77
 Corrosion 66, 67, 94, 95
 Corrosion resistance 35, 36, 37
 Coupling agents 71, 73
 Crack stopper 84
 Creep 5, 67
 Cristobalite 24
 Critical surface energy (CSE) 11, 69
 Cross-linking 9
 Crowns 59
 Curing *see* Polymerization
 Cutting efficiency 92, 93
- DBAs *see* Dentin bonding agents
 Dental amalgam 6, 66–67, 77, 94, 95
 Dental burs 92, 93
 Dental casting *see* Casting
 Dental cements 11, 54–55, 58–59, 60–61, 62–63, 79, 89
 Dental cutting 92–93
 Dental enamel 3, 6, 7, 12, 13, 62, 68, 69, 84, 93, 97
 Dental erosion 96–97
 Dental glass 81
 Dental implants 15, 40–41, 59
 Dental stone 18–19, 20–21
 Dental waxes 6, 22, 23
 Dentin 3, 6, 7, 12, 13, 62, 68, 69, 71, 84, 93
 Dentin adhesion 61
 Dentin bonding 63, 70–71
 Dentin bonding agents 62, 63, 70–71
 Dentin smear layer 13, 61, 70, 71, 74, 75, 93
 Dentinal tubules 12, 13
 Denture adhesives 64–65
 Denture base materials 44–45, 46–47
 Denture cleansing 51, 52, 53
- Denture fracture 48, 49
 Denture liner 9, 50, 51
 Denture plaque 53
 Denture repair 47, 48, 49
 Denture teeth 43, 45
 Dentures, complete (CDs) 41, 47, 49
 Dentures, partial *see* Partial dentures
 Devitrification 81
 Diametral tensile test 5
 Diamonds (diamond burs) 93
 Dibutyl phthalate (DBP) 45, 51
 Dicor 86, 87
 Die materials 19, 20, 21
 Die stone 19, 21
 Dielectric constant 6, 7
 Diffusive adhesion 9
 Digital imaging 89
 Dimethacrylate oligomer 73
 Direct bonding 69
 Dispersive adhesion 9
 Distractability 91
 DMFT score 13
 Drug abuse 97
 Dual-cure resin 63, 73, 77, 89
 Ductile materials 3
 Ductility 3
- EBA *see* Orthoethoxybenzoic acid
 EDP (2-ethylhexyl diphenyl phosphate) 51
 EDTA 70, 75
 Elastic deflection 91
 Elastic impression materials 30, 31
 Elastic limit 3
 Elastic materials 5
 Elastic modulus *see* Modulus of Elasticity
 Elastic recovery 30, 31
 Elastomer 3, 5, 47, 51
 Elastomeric chain 90, 91
 Electrical conductivity 6, 7
 Elgiloy 90, 91
 Elongation 34, 36, 38, 83
 Enamel *see* Dental Enamel
 Enamel bonding 69
 Endodontic irrigants 74, 75
 Endodontic materials 74–75
 Endosseous implants 41
 Enzyme cleaners 53
 Etching 67, 68, 69, 70, 71
 Eugenol 59, 76, 79
- Feldspar 80, 81
 Ferric chloride (FeCl₃) 63
 Fillers 45, 72–73, 84
 Film thickness 54, 57, 59, 60, 61, 62, 63
 Fineness 35
 Finishing 92–93
 Firing shrinkage 83
 Fixed partial dentures (FPDs) 36, 37, 59

Flexibility 31, 33, 43
 Flexural strength 42, 43, 85, 87
 Flow behavior 23, 29, 31, 55
 Flowable composite 77
 Fluidity 29, 37
 Fluorapatite 13, 69, 97
 Fluoridation 13, 69
 Fluoride release 57, 61, 62, 63
 Flux 81, 83
 Fracture toughness 5
 Free radical 44, 73
 Frit 81

 Gamma phase 66, 67
 Gantrez acid 65
 Gels 31
 GICs *see* Glass Ionomer Cement
 Giomer 63, 73
 Glass bridge 83
 Glass ionomer cement 6, 55, 57, 59, 60, 61, 77, 89
 Glass matrix 81
 Glass transition temperature, T_g 42, 44, 45, 51, 93
 Glass-ceramic 87
 Glazing 81, 82, 83
 GLUMA 63
 Glycol dimethacrylate 45
 GMPTS 73
 Gold alloys 6, 34, 35, 37, 39, 62
 Grinding 92–93
 Grossman's sealer 74
 Gum Arabic 23
 Gutta percha (GP) 75, 76–77
 Gypsum materials 18, 19, 20, 21
 Gypsum-bonded investments 24, 25

 Haptens 95
 Hard tissues 12, 13
 Hardening of metals 34, 35, 39
 Hardness 3, 19, 21, 31, 35, 38, 39, 43, 73
 Hardness, die materials 21
 Hardness, gypsum 19, 21
 Heat treatment of gold 34, 35, 37
 HEMA 62, 63, 71, 75
 Hemihydrate 19
 High copper amalgam 66–67
 High strength porcelain 84–85, 86–87
 Hooke's law 5
 Hue 7
 Hybrid layer 70, 71
 Hybrid resins 73
 Hydrocolloids 31, 65
 Hydrocryl® 47
 Hydrogen bond 9
 Hydrolytic breakdown 59
 Hydroxyapatite 12, 13, 14, 15, 41
 Hygroscopic effect 19
 Hyperalgesia 13
 Hypersensitivity 61, 94–95

 Imbibition 31
 Impact strength 42, 47
 Impact testing 42
 Implant metals 36, 40, 41
 Implants *see* Dental Implants
 Impression compound 28, 29
 Impression plaster 29
 Impression waxes 22, 23, 29
 InCeram® 84, 85, 86
 Indentation hardness 5
 Indium 67
 Inelastic impression materials 28
 Inlays and onlays 59
 Intaglio surface 47, 51, 65
 Interactive forces 11
 Interdigitation 9
 Interfacial effects 10, 11, 65
 Intracanal posts 77
 Investments 24, 25, 37
 Ion strengthening 85
 IRM® 77
 Irreversible hydrocolloid 30, 31

 Kaolin 80, 81
 Keesom forces 9
 Knoop hardness 3, 5, 47

 Lateral condensation 74, 75
 Leakage 71, 93
 Leucite 80, 81, 85, 87
 Light-cure resins *see* Photocuring resins
 Linear coefficient of thermal expansion 6, 7, 39, 81
 Liners 56–57, 50–51, 65
 Lining efficiency 6, 7
 Linkow blade vents 41
 Lithium disilicate 86, 87
 London forces 9
 Low strength bases 56–57
 Luting 54, 55, 89
 Luting agents *see* Dental cements
 Lyophobic conditions 11

 Malocclusion 91
 MAP (monoammonium phosphate) 25
 MDP 63
 Mechanical adhesion 10, 60, 65
 Mechanical properties 2–3, 4–5, 6–7, 23, 55, 57
 Mercaptan 31
 Mercury 6, 66, 67, 94, 95
 Metal allergy 43, 94, 95
 Metal casting shrinkage 25, 37
 Metallo-ceramic alloys 38–39
 Metamerism 7
 Methacrylate 62, 70, 73
 Methyl methacrylate 33, 44, 45, 49, 51, 62, 72, 95
 Microcracking 93
 Microleakage 74, 75, 77

- Micromechanical adhesion 11
- Microporosity 35, 68
- Mineral trioxide aggregate (MTA) 75
- Modeling plastic 29
- Modulus of elasticity 3, 31, 34, 35, 36, 38, 39, 43, 47, 56, 60, 72, 73, 85, 90
- Molybdenum 37
- Monomer *see* Methyl methacrylate
- Mousse 33
- Mucostatic action 29
- Munsell color system 7

- Nanostructurally integrating bioceramics (NIB) 61
- Newtonian fluid 29
- Nickel alloys 36, 37, 39, 95
- Nickel-chromium alloys 36, 37, 43, 62
- Nitinol (NiTi) 75, 90, 91
- Noble metal alloys 34, 35
- Non-eugenol cements 59
- Novus® 51
- NTG-GMA 63
- Nylon 43

- Obturation 74, 75
- Occlusal registration 32, 33
- Oligomer 47, 72
- Optical properties 7
- Orthodontic materials 63, 69, 90–91, 95
- Orthoethoxybenzoic acid (EBA) 29, 59, 77
- Osseointegration 15, 40, 41
- Osteoblasts 14, 15
- Osteoclasts 14, 15
- Osteocytes 15
- Osteoporosis 15
- OTC products 77

- P/L ratio 55, 59, 61
- Packable (condensable) resins 73
- Palatal vault 45
- Palladium 34, 35, 38, 39, 67
- Partial dentures 24, 36, 37, 42, 43, 47
- Passivation 36, 37
- Pattern resins 33
- Penetration coefficient 69
- PENTA 63
- Periodontal materials 78–79
- PFM restorations 9, 38, 39, 81, 84, 85
- Phosphate-bonded investment 24–25, 37
- Phosphoric acid 61, 69, 70, 71, 97
- Photocuring resins 44, 45, 46, 47, 49, 55, 57, 63, 73, 75, 76, 77, 79
- Photoinitiator 45, 47
- Plaster (of Paris) 19, 25, 29, 85
- Plastic deformation 4, 5
- Plasticization 45, 51,
- Plasticizer 45
- Platinum 35, 38, 39, 83

- PMDM 63
- PMMA *see* Poly(methyl methacrylate)
- Poisson's ratio 3
- Polishing 92–93
- Poly(ethyl methacrylate), PEMA 51
- Poly(hydroxymethacrylate) 47
- Poly(methyl methacrylate) 6, 33, 43, 44–45, 47, 49, 51, 57, 62, 63, 91, 95
- Poly[vinylmethylether maleate] 65
- Polyacrylic acid 61, 62, 63
- Polyamide 43
- Polyether 30
- Polymerization 9, 44, 45, 63, 73
- Polymerization shrinkage 30, 31, 33
- Polyphosphazene 50, 51
- Polysulfide rubber 6, 30, 31
- Polyvinylsiloxane (PVS) 6, 21, 31, 33
- Porcelain 6, 39, 54, 80–81, 82–83, 84–85
- Porcelain firing 39
- Porcelain-metal bonding 36, 38, 39
- Porosity 24, 35, 37, 67, 83
- Post-operative sensitivity 61, 69, 71
- Pour-type resins 47
- Precious metals 34, 35
- PREMA 51
- Primers 69, 70, 71
- Problems with denture bases 45
- Procera® 85
- Proportional limit 3, 34, 36
- Provisional cements 54, 58–59, 77, 79
- Provisional restorations/filling materials 47, 59, 76–77, 79
- Pulpal irritation 71
- Pulp-capping agents 57
- Pyroplastic flow 81

- Quartz (silica, SiO₂) 24–25, 63, 73

- Rapid-cure resins 47
- Refractories 25
- Removable partial denture (RPD) 36, 43, 47
- Residual monomer 45
- Resilience 3
- Resilient liner 45, 51, 53
- Resin cements 60
- Resin matrix 73
- Resin-modified glass ionomer cement (RMGI) 62–63, 89
- Resinomer 63, 73
- Resins 69, 70–71, 72–73, 74, 77
- Restoration leakage 71
- Rheology 11
- Rickert's sealer 74
- RMGI cement *see* Resin-modified glass ionomer cement
- Rockwell hardness 3
- Root canal therapy 74–75
- Rosin 23, 78

Saliva 5, 65, 79, 97
 Salt-and-pepper technique 49
 Sandblasting 68
 Sealing ability 77
 Self-cure resin *see* Auto-cure resin
 Self-etch primer 71
 Setting expansion 19, 21, 29
 Setting exotherm 77
 Setting shrinkage *see* Polymerization shrinkage
 Setting time 19, 21, 60, 62, 75
 Shear strength 5
 Shore hardness (Durometer) 3, 5, 21, 33
 Silane treatment 55, 63, 73
 Silica 24, 25, 65, 80, 81
 Silica filler 47, 63, 72–73, 77
 Silica-bonded investment 25, 37
 Silicate 80
 Silicone resin 51, 59
 Silver 67
 Sintering 9, 81, 82, 83, 85, 86, 87
 Slip casting 85, 87
 Slumping 81
 Smear layer *see* Dentin smear layer
 Sodium fluoride (NaF) 13, 69, 77
 Sodium silicate 80, 81
 Soft drinks 96, 97
 Soft liners *see* Resilient liners
 Solubility 54, 55, 57, 60, 62, 63
 Space maintainers 43, 91
 Spatulation 19
 Specific heat 6, 7
 Spinel 85, 87
 Springback 90, 91
 Sprues 24, 25, 35
 Stacking 83
 Stainless steel 36, 37, 90, 91, 95
 Stannous fluoride (SnF₂) 69
 Static fatigue 83
 Sticky wax 23
 Strain 3, 30, 31
 Stress 3
 Stress fatigue 87
 Stress intensity factor 5
 Stress relaxation 5, 23, 30, 31, 91
 Stress riser 5, 83
 Stress-strain curve 3, 4, 5
 Strontium 13
 Surface tension 11, 68, 74, 82
 Surgical pastes 29, 79
 Syneresis 31

 TEGDMA 63, 73
 Temporary cements *see* Provisional cements
 Temporary fillings *see* Provisional restorations
 Temporization 77
 Tensile strength 3, 5, 34, 36, 38, 42, 56, 60, 62, 67, 72
 Tensile strength, gypsum 19
 Tertiary amines 45, 71
 Tg *see* Glass transition temperature
 Thermal conductivity 6, 7, 38, 43, 55, 56
 Thermal diffusivity 6, 7
 Thermal effects 6, 23, 25, 56
 Thermal expansion 6, 7, 23, 24, 37, 39, 72, 91
 Thixotropy 11, 29, 31, 32, 33, 65
 Tin 67
 Tin plating 63
 Tissue conditioners 51, 53
 Titanium 15, 37, 39, 41
 Titratable acidity (TA) 97
 TMJ 43
 Topical fluorides 13
 Total etch technique 71
 Toughness 3, 43
 Toxicity 76, 77, 95
 Transverse strength 4, 5, 38, 42, 47, 49, 83
 Tray adhesives 31
 Triad® 47, 49
 Tribochemical bonding 55
 Trichromaticity 7
 Tridymite 24
 Tristimulus values 7
 Trituration 67

 UDMA 73
 Ultimate strength 3
 Urethane dimethacrylate (UDMA) 43, 47, 57, 63, 73

 Van der Waals forces 9
 Veneers 81, 87
 Vickers hardness 3, 34, 36, 38
 Vinyl polysiloxane *see* Polyvinylsiloxane
 Viscoelastic behavior 5, 31
 Viscosity 30, 31, 54, 55, 56, 57, 65, 67, 82
 Vitreous phase 81
 Vitrification 81, 83
 VLC bases 57
 VLC liners 57
 VLC resin *see* Photocuring resins
 VPS *see* Polyvinyl siloxanes

 W/P ratio 19, 21, 24
 Water solubility 72, 78, 79
 Waxes *see* Dental waxes
 Wear 5
 Wear resistance 62, 73
 Wetting 10, 65, 69
 Whitlockite 14, 15

 Xerostomia 65

 Yield point 3
 Yield strength 34, 36, 43, 91

Young's equation	11	Zinc oxide-eugenol (ZOE)	6, 28, 29, 33, 56, 57, 58, 59, 75, 77, 79
Young's modulus	<i>see</i> Modulus of elasticity	Zinc phosphate cement (ZNP)	6, 55, 57, 60, 77
Zinc	65, 66, 67, 95	Zinc polyacrylate/polycarboxylate (ZPC)	6, 9, 57, 59, 60, 61
Zinc oxide (ZnO)	59, 61, 77, 78, 79	Zirconia	86, 87

Keep up with critical fields

Would you like to receive up-to-date information on our books, journals and databases in the areas that interest you, direct to your mailbox?

Join the **Wiley e-mail service** - a convenient way to receive updates and exclusive discount offers on products from us.

Simply visit **www.wiley.com/email** and register online

We won't bombard you with emails and we'll only email you with information that's relevant to you. We will ALWAYS respect your e-mail privacy and NEVER sell, rent, or exchange your e-mail address to any outside company. Full details on our privacy policy can be found online.



www.wiley.com/email

